Odd Numbered Posters
Wednesday, March 11th 3:00 - 4:00pm

#	Title
1	**Distal Femoral Extension Osteotomies as Treatment of Knee Flexion Contracture in Patients with Cerebral Palsy** *Jason T. Rhodes, MD, MS, Robin Dorociak, BS, Molly Verschuyl, MPT, Ari Perlstein, BA, and Michael Aiona, MD*
3	**Conservative Versus Surgical Management of Knee Instability** *Natalie Johnston and John Yack*
5	**Long-Term Follow-up Surgical Impact on Different Segments of Clubfoot** *X.C. Liu, J. Thometz, R. Lyon, C. Tassone, and J. McKillip*
7	**10 Year Outcomes of Orthopedic Surgery in Persons with Cerebral Palsy: Soft Tissue Procedures** *Sylvia Õunpuu, MSc, Candace Nomides, DPT, Melany Westwell, MS, Katharine Bell, MS, Dawn Roller, PT, Matthew Solomito, BS, Dennis Tyburski, AS, Peter DeLuca, MD, and Elizabeth Weber, MD*
9	**Effects of Peroneal Nerve Stimulation During Gait for Persons with Multiple Sclerosis** *Stephanie Nogan Bailey, BSE, Lynne R. Sheffler, MD, and John Chae, MD*
11	**Case Study: Use of Botulinum Toxin in Charcot-Marie Tooth to Limit Muscle Activity in Gait** *Anna Murphy, PhD, Mary Danoudis, Dianne Cameron, and Barry Rawicki, MBBS*
13	**Classification of Mid-foot Break Using a Multi-segment Foot Model** *J.D. Maurer, A.H. Black, C.M. Alvarez, V. Ward, K.R. Davies, and R.D. Beauchamp*
15	**Use of Subtalar Joint Neutral for Static Offsets in Multi-Segment Foot Models** *Roy Wervey, Michael H. Schwartz, Sue Sohrweide, and Tom F. Novacheck*
17	**Gait Analysis of Patients with End-stage Ankle Osteoarthritis** *Michael E. Hahn, Elise S. Wright, Hannah Sutton, and Bruce J. Sangeorzan*
19	**Orthotic Control of Hyperpronation in the Rearfoot is Improved with Exercise** *David Hudson and James Richards*
21	**Efficacy of an AFO During Gait in Adults with Hemiplegia** *Karen J. Nolan, Mat Yarossi, Arvind Ramanujam, Krupa K. Savalia, Gail F. Forrest, and Elie P. Elovic*
23	**Ground Reaction Forces During Walking with Two Prosthetic Feet: A Preliminary Study** *Alison J. Behrens, Jeremy D. Smith, Jerome R. Rifkin, and Gary D. Heise*
25	**The Effects of Wearing a Functional Limb Length Adjustment Device with a Short-leg Walking Boot on Gait and Plantar Pressures: A Pilot Study** *J. Song, B. Heilman, C. Kihm, C. Nimick, P. McDonald, and E. Zoltick*
27	**Gait Analysis Post Hemipelvectomy/Sacrectomy with Spinopelvic Arthrodesis Reconstruction** *Krista A. ColemanWood, Michael J. Yaszemski, Franklin H. Sim, and Kenton R. Kaufman*

DISTAL FEMORAL EXTENSION OSTEOTOMIES AS TREATMENT OF KNEE FLEXION CONTRACTURE IN PATIENTS WITH CEREBRAL PALSY

Rhodes, Jason T., M.D., M.S.[2,3], Dorociak, Robin, B.S.[4],
Verschuyl, Molly, M.P.T.[5], Perlstein, Ari, B.A.[1,6], and Aiona, Michael, M.D.[1]
[1]Shriners Hospital for Children, Portland, Oregon, USA [2]The Children's Hospital, Denver, Colorado, USA [3]University of Colorado at Denver and Health Sciences Center, Denver, Colorado, USA [4]Northwest Biomechanics, Portland, Oregon, USA [5]Lake Washington Technical College, Kirkland, Washington, USA [6]Oregon Health Sciences Center, Portland, Oregon, USA

Introduction

Fixed knee flexion contracture can develop in the older ambulatory patient with cerebral palsy.[1] This makes gait inefficient with increased energy requirements and can cause anterior knee pain. There are two current treatments for fixed knee flexion contracture, serial casting or extension osteotomy, which can be combined with lengthening of contracted muscle-tendon units and restoration of the extensor mechanism to manage the dynamic flexed knee gait pattern present.[2] The purpose of this study was to report our results following distal femoral extension osteotomies in children with spastic cerebral palsy, specifically evaluating correction of the contracture and the effects on gait. A secondary analysis compared the results of an isolated osteotomy to those with additional soft tissue procedures including patellar advancement or a hamstring lengthening.

Statement of clinical significance

Distal femoral extension osteotomy is one option that is currently used to treat patients with cerebral palsy and knee flexion contractures. Evaluating the overall outcome of this surgery and when it is combined with other surgeries will aid in the treatment of patients with fixed knee flexion contracture.

Methods

This study is a retrospective cohort of 23 patients (43 sides, average age 14 ± 4 years) with spastic diplegic, hemiplegic, or quadriplegic cerebral palsy who had a fixed knee flexion contracture treated with distal femoral extension osteotomy. All patients included in the study had both pre and post operative physical exams, radiographic exams, and 3D computerized gait analysis. Physical exam data included motion about the hip, knee, and ankle. The dynamic gait data focused on knee extension and flexion kinematics and kinetics at selected positions in the gait cycle. PODCI data was also collected from a subset of these patients as it is not currently a required part or our gait analysis procedures. Twelve sets of data were available and were evaluated for patient and family satisfaction following surgery. A t-test was performed on all data to determine if significant differences existed.

Results

Significant results ($p<0.05$) are shown in Table 1 below. The entire group analysis showed significant improvement in the static contracture, which resulted in improvement in gait with less knee flexion in both stance and swing phase. The groups were separated into those with combined patellar advancement or a hamstring lengthening. These comparisons showed that there were significant beneficial changes in knee flexion and extension for both physical exams and kinematics for children who have had combined femoral osteotomies and patellar advancements. Patients improved an additional degree with the addition of patella advancement, and there was an improvement in knee position for combined hamstring lengthenings on kinematic data. Energy absorption decreased as the persistent knee flexion in stance decreased as did the demand on the quadriceps to maintain upright posture.

Variable	**Pre**	**Post**	**Change**
Physical Exam (N=41)	(degrees)		
Knee Flexion	140	130	-10
Knee Extension (Knee Flex Contracture)	-15	-3	12
Straight Leg Raise	56	61	5
Popliteal Angle	61	47	-14
Kinematics (N=41)	(degrees)		
Knee Flexion at Initial Contact	39	22	-17
Peak Knee Extension in Single Limb Stance	37	12	-25
Peak Knee Flexion in Swing	60	52	-8
Kinetics (N=27)	(Nm/Kg)		
Knee Peak extensor – EST	0.76	0.51	0.25
Knee Peak Flexor – MST	0.08	-0.05	0.13
Knee Peak extensor – TST	0.73	0.46	0.27
Knee Power Peak absorption – EST	-0.76	-0.41	0.35
Knee Power Peak absorption – TSW	-0.51	-0.67	0.16

Table 1: Significant overall results for physical exam, kinematics, and kinetics.

Data on patients who had a combined osteotomy and patellar advancement showed a significant increased knee extension on physical exam from a 17 degree contracture pre-operatively to 0 degrees post operatively compared to a change from a 13 degree to a 6 degree contracture for patients who underwent an isolated knee osteotomy. These patients also increased knee extension kinematically in single leg stance from 46 degrees flexed preoperatively to 7 degrees postoperatively as compared to a change from 29 to 16 degrees for isolated osteotomy patients. The PODCI data collected also showed that the families and patients had improvements in their overall satisfaction following this procedure.

Discussion

Distal femoral extension osteotomy when used on patients with cerebral palsy obtained a significant correction of fixed knee flexion contracture as measured by physical exam with minimal complications. There is increased knee extension and an associated decrease in knee flexion when this osteotomy is performed, which is seen on both physical exam and gait evaluations. This improved the kinematics and kinetics at the knee joint during ambulation. Combining this procedure with a patellar advancement resulted in further improvement and should be considered to improve gait further.

References

1. Rodda JM, et al. (2004), J Bone Joint Surg Br. 86:251-8.
2. Rodda JM, et al. (2006) J Bone Joint Surg Am. 88:2653-64

CONSERVATIVE VERSUS SURGICAL MANAGEMENT OF KNEE INSTABILITY

Natalie Johnston and John Yack
Gait Analysis Lab, University of Iowa Hospitals and Clinics, Iowa City, IA, USA

PATIENT HISTORY: The patient, a 41 year old female, was involved in an auto accident 21 years ago. She sustained a T6-T7 compression fracture, clavicle fracture, and partial meniscal tear in her right knee. Long-term recovery focused on chronic knee related issues, resulting in multiple knee surgeries over time with progressive development of arthrofibrosis (Table 1). She received physical therapy 3x/week off and on since the accident. This patient was referred to the University of Iowa Hospitals for orthopaedic consultation along with gait analysis and EMG prior to possible realignment osteotomy.

Table 1: Significant medical history detailed above.

1987	partial menisectomy
1988	scope debridement
1988	manipulation under anesthesia
1989	McKay Osteotomy
1990	posterior capsule release and notchoplasty
1991	ACL resection and hamstring release
1993	valgus-producing tibial osteotomy for medial compartment overload with over correction
1994	revision HTO varus producing to revive overcorrection
1995	HTO hardware removed
1996	referral to another surgeon
1997	arthroscopic lysis of adhesions and anterior interval release
1999	arthroscopic lysis of adhesions
1999	coronal plane medial & lateral, posterolateral, and ACL instability
2003	arthroscopic loose body removal, lysis of adhesions, and synovectomy
2005	MRI findings: scarred ACL, fibers remaining in continuity. Arthroscopic ACL healing response
2006	progressive perceived instability
2007	reports of knee giving way

CLINICAL DATA: AROM right knee 1-114 degrees, PROM 0-130 degrees. No significant hip or ankle deficiencies in active range of motion. Strength findings include right side hip extension (3+/5), hip adduction (3+/5), hip flexion (4/5), knee flexion (4/5). All other hip, knee, and ankle strength 5/5 including the left lower extremity. Long leg films showed neutral mechanical axis alignment, but single stance films showed a varus mechanical axis.

GAIT DATA: Patient typically used bilateral crutches to ambulate. Visual observation of gait shows an abnormal rapid thrust of her right knee into varus, then to valgus, then back to varus during stance phase.

3-D analysis, stance phase 0-75%: Sagittal plane kinematic patterns showed the right hip to be in flexion throughout the stance phase. The left knee was in hyperextension throughout stance. In the frontal plane, oscillations of the knee were documented with rapid movement into varus/valgus shortly after foot contact. In the transverse plane, hip abduction was associated with external rotation. The kinetics on the right side were generally decreased in amplitude, reflecting decreased weight support (max 60%) due to the use of a crutch on the left side. The oscillation seen in the frontal plane moment reflected relative motion of the ground reaction force in the frontal plane as the knee position changed.

EMG linear envelope data for the hip abductors and adductors did not indicate classic co-contraction to increase stiffness at the hip in the frontal plane during weight acceptance. EMG data did show a cyclic activation pattern which would more likely be associated with central control to govern the movement of the femur. When these EMG patterns are matched to the knee kinematic patterns in the frontal plane (Figure 1) the coordinated actions of these muscles aligned well with the motions at the knee (abductor activation being associated with lateral motion of the femur and a more varus position of the knee). It should be noted that during this interval the patient is bearing less than 25% of their weight through the right lower limb (as compared to 60% during the later part of stance).

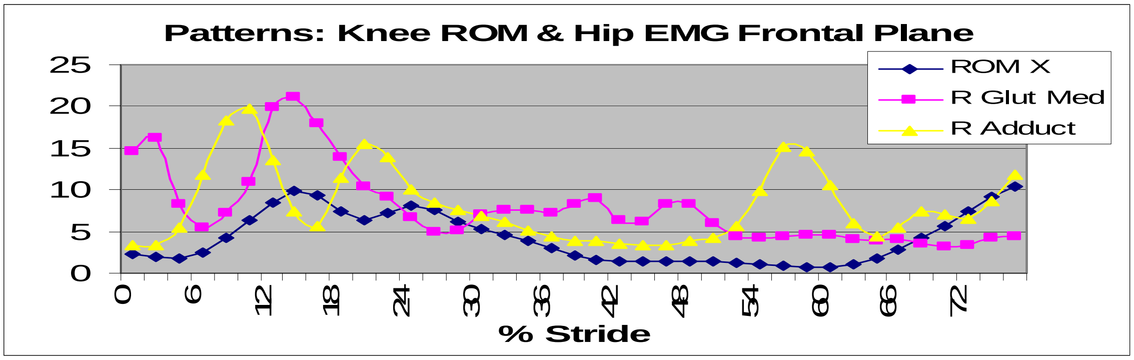


Figure 1: Overlay plot of knee motion in the frontal plane along with EMG of the hip muscles.

TREATMENT DECISIONS AND INDICATIONS: The knee control issue suggests two main possible explanations: 1) this patient has sufficient instability in her knee with poor proprioception and poor control of her muscles; when she weights her knee it falls into varus alignment rapidly causing a rapid response motion. 2) the rapid thrust and instability is due to central control rather than a distal mechanical problem. The EMG data suggest that the gait deviation may be a consequence to central control factors. The timing of hip muscle activation within the gait cycle was inconsistent with expected activation patterns through the stance phase and at times during the swing phase. The hip muscles appeared to create the varus then valgus then varus position of the knee by alternate activation of the abductors and adductors. She had fairly little quad and hamstring activity during this time. The cyclical muscle activation preceded the movement of the knee suggesting central control involvement rather than a reaction to an unstable position.

SUMMARY: Visual observation revealed an unstable gait, dependency on crutches and disability. During the oscillations of this patient's knee into varus, valgus, then back to varus, double stance phase was maintained bearing approximately 25% weight on her right knee; although she carries 60% of her weight later in stance phase. During the rapid oscillations, the right hip abductor and adductors fired sequentially. The timing of muscle activation and kinematic patterns were determined through motion analysis. Assessment of this patient through the use of clinical gait analysis led to the conclusion that this was conscious or unconscious voluntary knee motion. Following gait analysis with EMG, kinematic, and kinetic components, the data did not support proceeding with the original surgical re-alignment osteotomy plan. Motion analysis data was an integral component in the evaluation, understanding, and decision making for this patient.

Long-Term Follow-up Surgical Impact on Different Segments of Clubfoot

Liu, X.C.; Thometz, J.; Lyon, R: Tassone, C.; and McKillip, J*

Dept of Orthopedic Surgery, Children's Hospital of Wisconsin, Medical College of Wisconsin, Milwaukee, WI. *Dept. of Psychology, Southern Illinois University, IL

Introduction: With the development of a multi-segment foot model for objective analysis [1], it is possible to assess the function of a variety of foot deformities in adults. However, few case studies have been conducted evaluating multiple segment foot models in children with clubfoot[2]. The purpose of this study was 1) to have a long-term follow-up on clubfoot patients with post surgical status using the 7-segment-foot model; 2) to compare the differences in the foot model's kinematics between normal and clubfoot children.

Clinical Significance: Clubfoot clinical evaluation systems such as the Laaveg-Ponseti score, and the McKay score reflect subjective parameters based upon satisfaction, pain, and observational function. The 7-segment foot model may provide insight into the surgical outcome in children with clubfoot.

Methods: 26 patients with 45 clubfeet and 23 children with 46 normal feet were evaluated with a mean age of 11 years, 11 months. The mean follow-up for each patient was 10 years, 2 months. Subjects had physical and radiographic examinations, a dynamic foot analysis, clubfoot-specific instruments, and generic patient-based assessments. The dynamic foot analysis included the use of the 7-segment-foot model and a plantar pressure analysis.

The 7-segment-foot model consists of the tibia, calcaneus, navicular, cuboid, 1st metatarsal, 5th metatarsal and hallux. The patients had sensors attached and were asked to walk along a 20 ft walkway at a self selected pace. Each patient did three gait trials. The system enabled us to measure movement bilaterally using the 7-segment-foot model. Motion was sampled at 120Hz using the StarTrak electromagnetic tracking system (Polhemus Inc., Colchester, VT) (ETS). Differences between children with clubfoot and normal populations were tested through analysis of variance (ANOVA) using SPSS (Statistical Program for the Social Sciences, 11.0, Chicago, IL). A discriminate analysis was performed to determine the important predictors differentiating a clubfoot from a normal foot.

Results: Five kinematic variables were identified as significantly different between the normal and clubfeet by the Stepwise discriminate analysis using the 7-segment model (Table 1and Figure 1).

Table 1. Significant differences between normal and clubfeet children ($P<0.05$)

Kinematic Variables	**Normal**	**Clubfoot**	**P-Value**
Ankle plantarflexion ST	-11.2±4.6	-6.8±3.9	<0.0001
Ankle Supination SW	6.9±4.1	4.4±3.9	0.003
1stMet.-Hallux Max.dorsiflexion SW	34.8±9.5	17.5±8.2	<0.0001

1stMet.-Hallux Max.supination SW	7.1±5.3	4.1±6.8	0.02
Nav.-1st Met.Max.dorsiflexion SW	-1.8±3.4	2.4±5.7	<0.0001

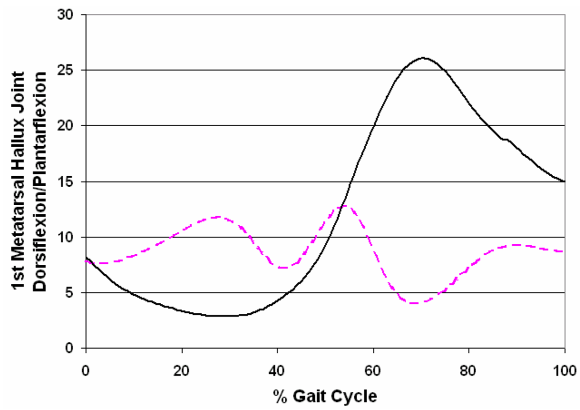


Figure 1. Comparison of dorsiflexion/plantarflexion of 1st metatarsal-hallux joint in one patient between normal foot (solid line) and clubfoot (dot line).

Discussion: Although clubfoot clinical evaluation systems (Laaveg, and McKay etc) indicated significant differences of the overall scores between clubfoot and normal foot, they were unable to determine the degree or the location of the foot deformities. Our current 7-segment-foot model reliably provides a 3D kinematic explanation of these deformities, and successfully detects a correction resulting from a completed subtalar release procedure in children with clubfoot.

References:

1. Leardini A, Benedetti MG, Catani F, Simoncini L, and Giannini S. An anatomically based protocol for the description of foot segment kinematics during gait. Clinical Biomechanics 1999, 14:528-536.
2. Khazzam M, Roh JY, Long JT, Hassani S, Harris GF, Graf AN, and Kuo K. Functional gait analysis in children following clubfoot release. Foot and Ankle Motion Analysis. Editor, G.Harris etc. CRC Press, 2008

Acknowledgements:

This study was partly funded by the Orthopaedic Research and Education Foundation.

10 YEAR OUTCOMES OF ORTHOPEDIC SURGERY IN PERSONS WITH CEREBRAL PALSY: SOFT TISSUE PROCEDURES

Sylvia Õunpuu MSc, Candace Nomides DPT, Melany Westwell MS, Katharine Bell MS, Dawn Roller PT, Matthew Solomito BS, Dennis Tyburski AS, Peter DeLuca, MD and Elizabeth Weber MD
Center for Motion Analysis, Connecticut Children's Medical Center, Farmington, CT

Introduction

Multilevel surgical intervention is a common approach to the surgical correction of gait abnormalities in children with cerebral palsy (CP). The short-term outcomes for the combination of rectus femoris transfer, hamstring lengthening and gastrocnemius lengthening have been well documented using three-dimensional motion analysis techniques[1]. However, the impact of time, growth, and puberty on these short-term outcomes for this common combination of procedures is not well understood. Therefore, the purpose of this study was to evaluate the long-term effects of these same soft tissue procedures on gait in patients with CP.

Statement of Clinical Significance

Studying long-term outcomes of surgical intervention will allow clinicians to 1) understand long-term benefits and sequela of surgical procedures 2) better estimate ambulatory prognosis for their patients and 3) ultimately allow for more critical surgical decision-making which will improve outcomes.

Methods

Forty persons with CP volunteered to participate in this study. A subgroup of 20 subjects (n=33 sides) underwent rectus femoris transfers, medial hamstring and gastrocnemius lengthenings in combination with a selection of other soft tissue and/or bony procedures at the foot and hip. Each subject underwent a pre-operative (P0) and post-operative (P1) gait analysis study before and 1-year after orthopedic surgery as part of the routine clinical decision making process which included the following procedures: physical examination, video and motion analysis[2]. Subjects were then recruited to participate in a second post-operative gait analysis study (P2). Subjects were excluded if they had any intervening surgery between P1 and P2. Kinematic, kinetic and clinical variables collected at P0, P1 and P2 were compared using a repeated measures analysis of variance (ANOVA) and Duncan's Multiple Range post-hoc testing.

Results

The average age at P0 was 9±4, P1 was 10.6±4 and P2 was 20±4 years. The P2 was a mean of 11±2 years after surgery. Results show initial improvements in passive range of motion measures are not sustained over time, however, gait changes in many cases are maintained over time (Tables 1 and 2).

Table 1: A comparison of selected mean (±1 S.D.) P0, P1 and P2 clinical examination and kinematic parameters for the ankle. ([1] significant difference between P0 and P1, $p<0.05$; [2] significant difference between P0 and P2, $p<0.05$; [3] significant difference between P1 and P2, $p<0.05$; g.c. = gait cycle; deg = degrees)

	Dorsi-flexion knee 0 (deg)	Dorsi-flexion knee 90 (deg)	Ankle angle initial contact (deg)	Peak dorsi-flexion stance (deg)	Time peak dorsi-flexion stance (%g.c.)	Peak plantar flexion swing (deg)	Ankle range of motion (deg)
P0	2±14	10±12	-9±14	1±19	23±19	-24±19	26±12
P1	15±9[1]	21±7[1]	1±5[1]	15±7[1]	40±19[1]	-5±8[1]	22±6[1]
P2	-3±10[2,3]	10±8[3]	-1±6[2]	15±7[2]	44±10[2]	-4±6[2]	20±5[2]
typical			-1±5	14±3	34±9	3±4	32±7

Table 2: A comparison of selected mean (±1 S.D.) P0, P1 and P2 clinical examination and kinematic parameters for the knee. ([1] significant difference between P0 and P1, $p<0.05$; [2] significant difference between P0 and P2, $p<0.05$; [3] significant difference between P1 and P2, $p<0.05$; g.c. = gait cycle; deg = degrees)

	Straight leg raise (deg)	Popliteal angle (deg)	Knee angle initial contact (deg)	Mean flexion knee stance (deg)	Peak knee extens mid stance (deg)	Peak knee flexion swing (deg)	Peak knee flexion swing (%g.c.)	Knee range of motion (deg)
P0	52±10	-51± 14	34±9	23±10	13± 12	56±11	82±7	43±15
P1	63± 10[1]	-32±10[1]	22±8[1]	18±10	9±11	55±11	77±5[1]	47±13
P2	51±13[3]	-54±13[3]	23±9[2]	19±10	10±11	47±9[2,3]	77±6[2]	38±9[3]
typical			9±6	15±5	6±6	66±7	71±2	63±7

Discussion

Significant changes in both clinical and gait variables from pre to 1 year post surgery indicate the short-term gait benefits of the combination of rectus femoris transfer, hamstring lengthening and gastrocnemius lengthening procedures. At an average of 11 years following surgical intervention, passive range of motion gains noted at 1 year after surgery were lost at the knee and ankle, however, certain gait changes are maintained. At the ankle, the peak dorsiflexion in stance is maintained over 11 years; however the time to peak dorsiflexion is delayed. This may be due to increasing functional weakness of the ankle plantar flexors relative to body weight over time as ankle plantar flexor strength on clinical assessment remained unchanged. At the knee, improvements in extension at initial contact are maintained over the long-term; however, mean knee flexion in stance is not changed. Peak knee flexion in swing shows a decline over time, which is similar to a non-operative comparison group over a 4 year interval[3]. The reason for this decline is unclear but may be related to increasing "stiffness" over time experienced by many persons with CP. In the future, patients will be divided into preoperative knee kinematic patterns to assist in the assessment of outcomes based on preoperative knee function.

References 1. Adolfsen S, Õunpuu S and DeLuca PA *J Pediatr Orthop*, 2007, 27(6):658-667. 2. Davis R. et al. *Human Motion Analysis: Current Applications and Future Directions.* Piscataway: IEEE Press; 1996: 17-42. 3. Bell K, et al., *J Pediatr Orthop*, 2002, 22:677-682.

Acknowledgment: Funding provided by the Harold and Rebecca H. Gross Foundation

Effects of Peroneal Nerve Stimulation during Gait for Persons with Multiple Sclerosis

Stephanie Nogan Bailey, B.S.E.[1,2], Lynne R. Sheffler, M.D.[1,3], John Chae, M.D. [1,3]

1 - Cleveland Functional Electrical Stimulation Center
2- Louis Stokes Cleveland Department of Veterans Affairs Medical Center
3 - Case Western Reserve University and MetroHealth Medical Center

Introduction

The rehabilitation intervention considered standard of care for ankle weakness in multiple sclerosis (MS) is an ankle-foot-orthosis (AFO) [1], which holds the ankle in a neutral position for a safe heelstrike, but inhibits residual ankle motor movement. Functional electrical stimulation (FES), in the form of a peroneal nerve stimulator (PNS) [2], may be a promising alternative to the use of an AFO. With FES, intact lower motor neurons in paralyzed or paretic muscles are activated in a precise sequence to create muscle contractions to accomplish functional tasks [3].

Clinical Significance

Lower extremity motor impairment secondary to pyramidal tract involvement in MS is an early [4] and significant contributing factor to gait dysfunction. In chronic MS, motor impairment is a central feature in approximately 80% of all patients [5]. Although weakness in the tibialis anterior is often present [6], isometric dorsiflexion (DF) endurance exercises can improve walking ability [7]. A PNS to augment volitional DF is an appropriate alternative to an AFO for ankle stability and positioning during gait. To date, no direct comparison of the biomechanical effect of an AFO versus PNS has been performed. This knowledge would be critical to clinical decision making, as there are no guidelines as to who would benefit from which device.

Methods

Four subjects diagnosed with MS (> 6 months), ankle DF strength ≤ 4/5, and prior usage of a physician-prescribed AFO completed a minimum of 4 weeks of daily usage with a PNS for ambulation before undergoing quantitative gait analysis (QGA). QGA was performed under 3 test conditions in the following order: 1) no device (ND), 2) AFO and 3) PNS. The order of the testing conditions was purposefully not randomized to eliminate the concern that the ND or AFO performance might have been enhanced by a "carryover" effect of prior application of the PNS.

Retro-reflective markers were adhered to the skin at the anatomical locations for the lower extremity Plug-In-Gait model. Subjects ambulated along a 10 m walkway at their self-selected speed in view of the cameras of the Vicon 370 system. The following subset of spatio-temporal and kinematic parameters were compared for the affected limb under the 3 testing conditions: walking speed (m/s), peak pelvic obliquity and knee flexion during swing (deg), and ankle angle at initial contact (IC) (deg). The global variable speed and a peak angle for each joint were selected to give a representation of the subjects' walking patterns.

Results

Data are presented as a case series in the table below. A single asterisk (*) signifies that the PNS demonstrated a statistically significant improvement ($p < 0.05$) over the ND or AFO conditions, while a double asterisk (**) signifies that the PNS demonstrated a statistically significant improvement ($p < 0.05$) over both the ND and AFO conditions.

Walk Speed	S1		S2		S3		S4	
(m/s)	Mean	StDev	Mean	StDev	Mean	StDev	Mean	StDev
ND	1.01	0.02	0.23	0.03	0.84	0.08	0.92	0.33
AFO	0.96	0.04	0.22	0.04	0.80	0.05	1.00	0.03
PNS	1.05 *	0.01	0.27 **	0.03	0.79	0.05	0.95	0.06
Pelv Obliq Sw	S1		S2		S3		S4	
(deg)	Mean	StDev	Mean	StDev	Mean	StDev	Mean	StDev
ND	-2.22	1.52	1.62	1.44	.4-97	2.63	-2.75	0.63
AFO	-3.09	1.17	3.37	2.33	-5.11	1.26	-1.44	0.82
PNS	-5.04	2.32	1.54	0.97	-6.18	1.08	-3.82	0.83
Knee Flex Sw	S1		S2		S3		S4	
(deg)	Mean	StDev	Mean	StDev	Mean	StDev	Mean	StDev
ND	49.01	2.68	32.46	3.38	57.13	2.76	34.61	2.32
AFO	55.51	3.79	32.02	5.15	55.46	1.80	35.61	2.33
PNS	63.43 **	2.39	37.02 *	2.18	57.18	2.67	35.93	2.58
Ank Ang at IC	S1		S2		S3		S4	
(deg)	Mean	StDev	Mean	StDev	Mean	StDev	Mean	StDev
ND	5.36	2.13	2.26	1.40	7.37	1.46	2.62	3.83
AFO	6.15	0.62	9.93	0.57	8.13	0.49	3.51	3.18
PNS	8.32 **	1.48	20.84 **	1.40	10.56 **	0.46	4.96	4.08

Discussion

The PNS caused variable, but notable, effects on the gait of each of the 4 subjects. In general, a flexor-synergy pattern appeared and improved the clearance of the limb during swing phase. Future investigations may discover which clinical characteristics would benefit the most from the use of FES. QGA is an important and useful tool in measuring the effects of the PNS in comparison to traditional bracing and therefore QGA could be used to make clinical decisions.

References

1. Cobble, D., et al. Rehabilitation of the Patient with Multiple Sclerosis, 1993, p861-885.
2. Lyons, G.M., et al. IEEE Trans Neural Systems and Rehab Engineering, 10(4): 260-79, 2002.
3. Sheffler, L.R., Chae J. Muscle Nerve, 35(5): 562-590, 2007.
4. Martin, C.L., et al. Multiple Sclerosis, 12(5): 620-8, 2006.
5. Matthews, B. Symptoms and Signs of Multiple Sclerosis in McAlpine's MS, 1998, p145-190.
6. Ng, A.V., et al. Muscle Nerve, 29(6): 843-52, 2004.
7. Mount, J., Dacko S. Neuro Rehabilitation, 21(1): 43-50, 2006.

Acknowledgements

This work was supported by the Ohio Neurostimulation and Neuromodulation Partnership, Ohio Board of Regents, Biomedical Technology Transfer Award.

CASE STUDY: USE OF BOTULINUM TOXIN IN CHARCOT-MARIE TOOTH TO LIMIT MUSCLE ACTIVITY IN GAIT

Murphy, Anna, PhD[1,2], Danoudis, Mary[1], Cameron, Dianne[1,2], Rawicki, Barry, MBBS[1,2]
[1]Kingston Centre, Southern Health, Melbourne, Australia, [2]Monash University, Australia

Patient History

Sixteen year old girl (CS) diagnosed with Charcot-Marie Tooth syndrome (CMT), Type 1A, at 12 years of age. She walked independently but distance walked was limited by fatigue and painful feet. She reported difficulty in standing and fell once or twice a week due to trips. She undertook a rehabilitation program at 15.5 years of age which included exercises to improve joint range of motion and muscle strength. Bilateral orthotics with heel raises were prescribed to improve standing balance, and night splints to increase calf range. She was referred to the Clinical Gait Analysis Service to assist in determining appropriate treatment for her marked equinus and knee hyperextension which were having a significant impact on her gait.

Clinical Data

On examination CS had significant weakness of her left and right dorsi flexors and moderate weakness of the left and right peronei muscles. She had significant passive tightness of the left and right gastrocnemius and soleus with no dynamic tightness evident (Table 1).

Muscle Group	Range of Motion (°)		Strength (MRC Grades)	
	Left (R2)	Right (R2)	Left	Right
Dorsiflexion (K@0°)	-15	-12	1	1
Dorsiflexion (K@90°)	-8	-12	1	1
Evertors	½ range	½ range	3	3

Table 1: Key range of motion and strength data measured on clinical assessment

Gait Data

CS was assessed walking in bare feet and with her shoes and orthotics at her preferred speed. Gait analysis showed both ankles remained in plantar flexion during stance due to bilateral calf tightness. The impact on her balance was partly compensated by increasing anterior pelvic tilt. The substantially reduced range in the left calf muscles was associated with knee hyperextension, a large knee flexor moment and increased power absorption at the knee and ankle during early stance. Both ankles were in excessive plantar flexion during swing, which was likely to be due to weak dorsi flexors and increased inversion. In shoes with orthotics, pelvic tilt was reduced and the left knee was no longer hyperextended, which corresponded with significant improvements in the moments and powers at this joint. There was little change in the ankle kinematics but the power absorption and generation in early to mid stance was decreased, which was likely to be due to the decreased stretch on the calf muscles.

Treatment Decisions and Indications

Though there is very little evidence of use of Botulinum toxin (BoNT-A) injections in CMT, the clinical team recommended injections to the tibialis posterior bilaterally in an attempt to improve foot position by decreasing ankle plantar flexion and inversion. The injections were to be immediately followed with serial casting to improve muscle length of the gastrocnemius and soleus. Given the improvement with the shoes and orthotics, dynamic ankle foot orthoses (AFOs) were prescribed to control plantar flexion at initial contact and swing, and medial-lateral stability at mid-stance.

A secondary gait analysis was performed six weeks post injections and five weeks of serial casting. The patient reported the foot pain had subsided and balance had improved. On clinical examination, passive range of dorsi flexion motion had increased by 12° and 13° respectively for the right and left soleus. Dorsi flexors were more readily isolated and strength had improved to 3-4 (MRC Grade).

Gait data indicated marked improvement, with bilateral dorsi flexion and minimal knee hyperextension through mid to late stance phase (Figure 1). With an increased walking speed, peak joint moments and powers were increased; but the shapes were closer to normal comparison data. The dynamic AFOs, although improving ankle kinematics, increased instability about the hip and knee joints which may have reflected the short time she had been using the AFOs.

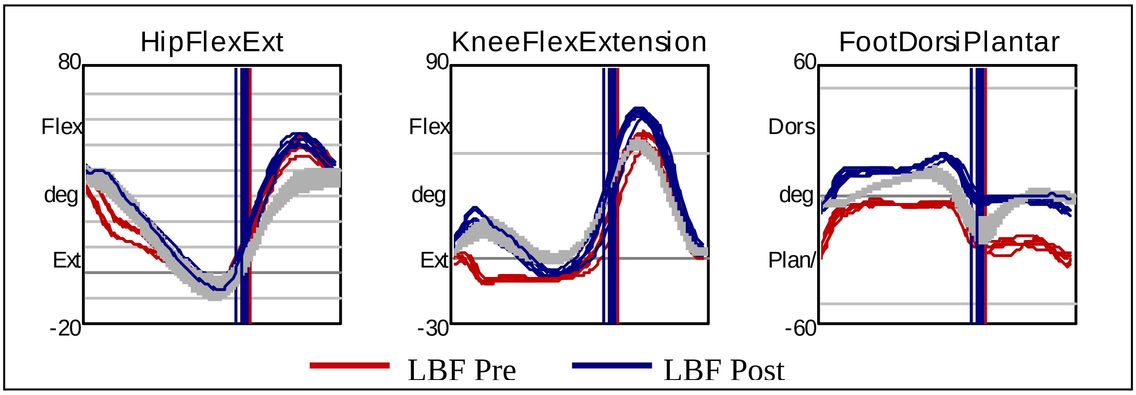


Figure 1: Left hip, knee and ankle joint kinematics pre and post BoNT-A injections and serial casting

CS was reviewed six months later when she reported her heels were further off the ground and standing balance had deteriorated. Clinical examination showed a 15° increase in passive dorsiflexion tightness on the left but no loss of muscle strength. Left dorsiflexion range had also decreased during gait which was associated with an increase in knee hyperextension. There was only a small deterioration on the right side. Further BoNT-A injections and serial casting were recommended following this assessment.

Summary

This case study highlights how gait analysis was a key component in the evaluation and understanding of treatment outcomes in a teenager with CMT. The use of BoNT-A and serial casting to reduce activity of unopposed muscle groups during gait and lengthen the calf muscles significantly improved CS's gait and ability to develop force through mid-range. This treatment also reduced her pain. The capacity to maintain these improvements is evidenced, and even though there was deterioration in the most affected side, on-going treatment may potentially limit the effect of CMT disease on gait.

Classification of Mid-foot Break Using a Multi-segment Foot Model

J.D. Maurer, A.H. Black, C.M. Alvarez, V. Ward, K.R. Davies, R.D. Beauchamp
Shriners Gait Lab, Vancouver, BC, Cananda
Children's & Women's Health Centre of British Columbia, Vancouver, BC, Canada

Introduction: The foot is composed of multiple joints controlled by both intrinsic and extrinsic muscles. If any of these joints exhibit restricted movement, compensatory adaptations may occur at one or more of the other joints. For example, with reduced dorsiflexion of the ankle, spurious dorsiflexion of the foot may be seen to occur through the mid-foot (either medially or laterally) while eversion may also occur through the mid-foot. This may lead to the development of a "mid-foot break" (MFB).
Researchers have recently begun to examine joint motion within the foot during normal gait. Traditionally, the foot has been modeled as a single rigid segment for the purposes of gait analysis. With the advancement of high resolution cameras and overall improved technology, researchers and clinicians are now able to record the position of much smaller markers at higher frame rates. This has led to the development of several multi-segment foot models[1-4].

Clinical Significance: Several studies can be found in the literature describing kinematic patterns of the forefoot (FF) and hindfoot (HF) during normal gait[1-4], however, few have described dynamic mid-foot motion in children with pathological gait patterns. In particular, no studies have examined foot kinematics in children who have developed a MFB. Prevention of the development of a MFB in neurological conditions such as CP usually revolves around maintaining efficient lever-arm integrity to generate sufficient forces for an effective plantarflexed toe off. Studying the FF and HF kinematics in this population may lead to earlier identification of MFB, and assist with preventative treatments such as Botox injections or surgery. The purpose of this study was to characterize the kinematic patterns of FF and HF motion during gait in children with MFB.

Methods: Study participants were divided into 2 groups: (a) children with unilateral or bilateral mid-foot break as determined by an orthopedic surgeon or registered physiotherapist (Left: n = 7, Right: n = 13), and (b) children with no evidence of mid-foot break or any gait abnormalities (Left: n = 7, Right: n = 7). An eight-camera Motion Analysis system (Motion Analysis Corporation, Santa Rosa, CA) was used to record the 3-dimensional positions of reflective markers placed according to the conventional Helen Hayes marker set. Markers were also placed on the HF and FF following guidelines for the Oxford Foot Model[1,2]. Three complete stride cycles were recorded for each subject on the affected side(s). Foot motion was evaluated in two different ways: (1) motion of the foot modeled as a single rigid segment, and (2) motion of the HF and FF modeled as separate rigid segments. In both cases, the motion was calculated relative to the lab coordinate systems (LCS) and relative to the proximal segment.

Results: The kinematic data in the sagittal plane demonstrated a marked difference between children with and without mid-foot break (Figures 1 & 2). The single-segment foot model indicated an absent first rocker, reduced peak dorsiflexion (DF) during stance, and a plantarflexed (PF) foot position during swing in the MFB group. When the foot was examined as two segments, those with MFB demonstrated a plantarflexed HF throughout both stance and swing phase. Relative to the HF, the FF maintained a dorsiflexed position throughout stance and swing. Peak FF dorsiflexion occurred at 46% of the gait cycle in both groups, with a mean value of 18.0° ± 9.7° DF in the MFB group and 5.5° ± 4.5° DF in the Normal group. Relative to the LCS, the foot position at initial contact was 1.3° ± 10.5° DF in the MFB group compared to 19.5° ± 4.9° DF in the Normal Group. Between 10-40% of gait cycle, the MFB group had a slightly more plantarflexed foot position relative to the LCS compared to Normal (mean diff = 5.8° ± 1.0°). During the same period, the FF remained parallel to the ground in both groups (mean FF position = 3.0° ± 2.2° PF for MFB; 1.0° ± 1.5° PF for Normal); however, compared to the Normal group, the MFB group demonstrated greater HF plantarflexion (mean diff = 13.7° ± 0.6°) relative to the LCS.

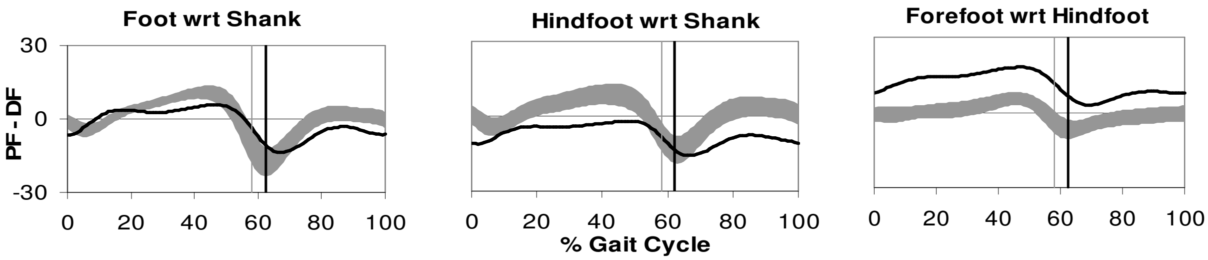


Figure 1: Foot, HF and FF Angles (degrees) calculated with respect to the Proximal Segment. Normal mean ± stdev shown as grey band. MFB mean shown as solid black line.

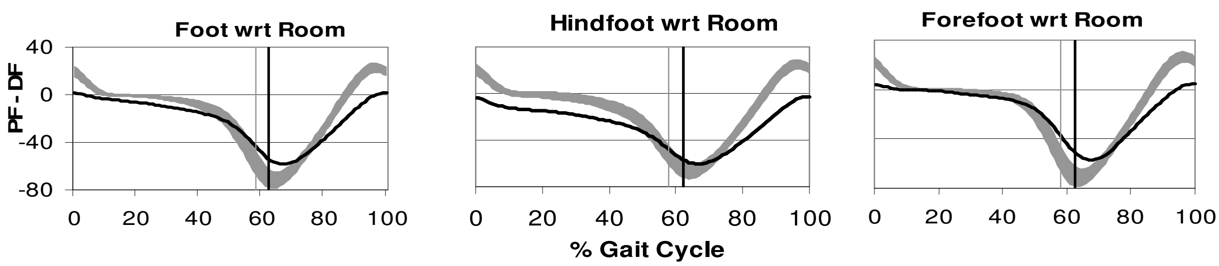


Figure 2: Foot, HF and FF Angles (degrees) calculated with respect to the LCS. Normal mean ± stdev shown as grey band. MFB mean shown as solid black line.

Discussion: Comparing the data from the conventional foot model with the multi-segmental foot kinematics, it can be seen that the conventional foot model mis-represented the true foot motion during stance phase by underestimating the equinus across the ankle. In the MFB group, the HF continued to plantarflex throughout stance phase, while the FF remained in full contact with the floor until approximately 40% of stance phase. By using this model it is now possible to quantify the presence and severity of MFB in children in the sagittal plane.

References:

1. Stebbins et al. Gait Posture 23(4): 401-10, 2006.
2. Carson et al. J Biomech 34:1299–307, 2001
3. Kidder et al. IEEEE Trans Rehab Eng 4:25-32, 1996.
4. Davis et al. Foot and Ankle Motion Analysis, 425-444, 2008.

Acknowledgements: This work was supported by an unrestricted grant from Allergan Canada

Use of Subtalar Joint Neutral for Static Offsets in Multi-Segment Foot Models

Roy Wervey[1], Michael H. Schwartz [1,2], Sue Sohrweide[1], Tom F. Novacheck[1,2]
[1] Gillette Childrens Specialty Healthcare, St. Paul USA
[2] University of Minnesota, Minneapolis, USA

INTRODUCTION

Determining an appropriate kinematic reference is one of the major challenges in foot modeling. Many foot models use a weight bearing (WB) static trial to establish reference angles [1,2]. Using offset angles from a WB static, however, can have the effect of masking foot deformity. For example, a subject with a fixed hind foot valgus in both static and dynamic conditions will appear to have a neutral hind foot during gait. The subtalar joint neutral (STJN) position provides non weight-bearing foot alignment reference position using a standardized method [3]. A weight bearing STJN position has been used for reference in the past, however for subjects with severe foot, general orthopaedic, and neuromuscular pathology, this approach may not be useful or practical [4]. Previous work in our lab using a three-dimensional multi-segment foot model has established that the inter-rater reliability of the hind-foot position relative to the shank in STJN position is good [5]. Using motion capture to measure fore-foot position in STJN is challenging because the patient must lie prone while the examiner positions the foot and reflective markers are easily obscured. The purpose of this study was to examine the feasibility of collecting static foot model data in a non weight-bearing STJN position.

CLINICAL SIGNIFICANCE

Establishing a meaningful kinematic reference is important when evaluating data obtained from multi-segment foot models. For patients with foot deformity, using WB static offset angles can mask both the fixed and compensatory components of foot deformity. The STJN static position provides both a reference that is both standardized and clinically meaningful.

METHODS

A Vicon MX 12 camera system was used to collect motion capture data for a three-dimensional multi-segment model foot model [1]. To examine the feasibility and quality of this process a STJN static, WB static, and walking data were collected for two adult volunteers. An adjustable portable plinth was used to optimally position the subject's foot and shank in the motion capture volume while an experienced examiner placed the foot in STJN. Several seconds of data were then collected, and joint rotation angles (offset angles) for the WB and STJN statics were calculated for the calcaneus relative to shank, and forefoot relative to calcaneus. These offset angles were then subtracted from the walking trial joint rotations to yield offset corrected plots.

Figure 1. STJN static motion capture setup.

RESULTS

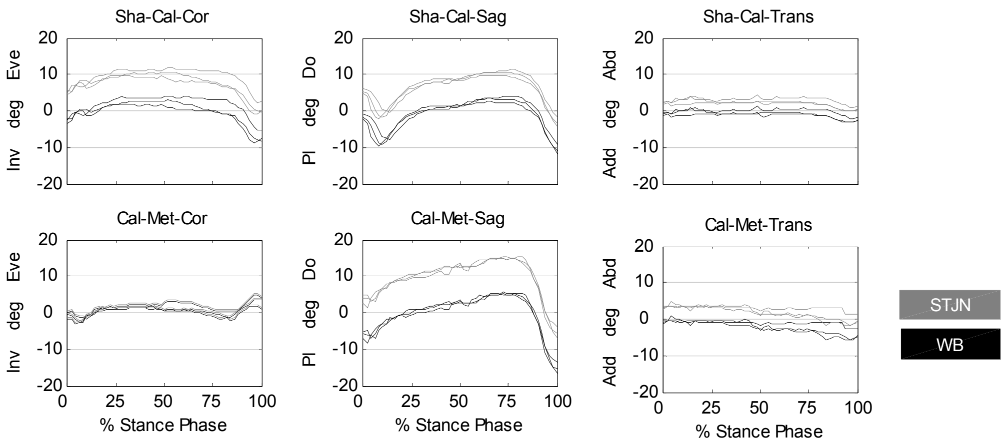


Figure 2. Sample data comparing WB vs. STJN offset angle corrected kinematics: calcaneus relative to shank, and forefoot relative to calcaneus.

Table 1. Difference between WB and STJN offset angles in degrees (absolute value).

	Subject 1		Subject 2	
	Left	Right	Left	Right
Shank vs. Calcaneus (coronal)	0.9	5.2	7.8	8.2
Shank vs. Calcaneus (transverse)	3.1	0.4	3.1	3.5
Shank vs. Calcaneus (sagittal)	9.5	1.1	7.2	14.9
Calcaneus vs. Metatarsus (coronal)	1.6	2.7	0.4	3.7
Calcaneus vs. Metatarsus (transverse)	1.5	2.4	0.3	0.3
Calcaneus vs. Metatarsus (sagittal)	6.1	6.6	1.9	7.2

DISCUSSION

By carefully positioning our subjects in the motion capture volume we were able to reliably obtain kinematic offset angles with the foot positioned in STJN. Markers on the front of the shank were difficult for the motion capture system to reconstruct and may be replaced with virtual markers in the future. There were marked differences between STJN and WB static offset values even for typical feet; we expect these differences to be larger in feet with deformity. In order to establish clinical utility, the reliability STJN position for the forefoot relative to calcaneus must be assessed, as well as the practicality of the method with pathological feet.

REFERENCES

1. Leardini A, *et al*. (2007) Gait Posture, **25**:453-62.
2. Stebbins J, *et al*. (2006) Gait Posture, **23**:401-10
3. Root M.L., et al. <u>Biomechanical Examination of the Foot</u>, Clinic. Biomech. Corp. 1971.
4. Houck J, et al (2008) Gait Posture, **28**:29–37
5. Elli K, *et al*. (2008) *Proceedings* GCMAS 13th Annual Meeting:208-209

ACKNOWLEDGEMENTS:

We would like to thank Kristen Elli for her implementation of the IOR foot model at Gillette Childrens Specialty Healthcare Center for Gait and Motion Analysis.

GAIT ANALYSIS OF PATIENTS WITH END-STAGE ANKLE OSTEOARTHRITIS

Michael E. Hahn[1,2], Elise S. Wright[1], Hannah Sutton[1], and Bruce J. Sangeorzan[1,3]
[1]RR&D Center of Excellence, VA Puget Sound, Seattle, WA; [2]Department of Mechanical Engineering, [3]Orthopaedics and Sports Medicine, University of Washington, Seattle, WA

INTRODUCTION

Osteoarthritis (OA) of the ankle affects 6-10% of the population [1]. There is evidence that the incidence of ankle OA is increasing [2,3]. Relatively few studies have investigated the functional limitations of ankle OA during gait; reporting decreased range of motion, prolonged stance time, shortened stride length, reduced cadence and preferred walking speed, lower sagittal and transverse plane moments, and reduced ankle power in patients with ankle OA [4,5]. It has been suggested that patients with ankle OA alter gait patterns to compensate for pain and reduced ROM in the arthritic joint; however, this notion has yet to be tested. Furthermore, no study has examined gait alterations in all joints of the affected limb. The purpose of this study was to determine the gait characteristics of a sample of patients with end-stage ankle OA. We hypothesized that patients would have reduced temporal-distance (TD) magnitude, reduced hip, knee and ankle ROM, and altered joint kinetics when compared between limbs (affected [AFF] vs. unaffected [UN]) and with matched controls (CON).

CLINICAL SIGNIFICANCE

The US Department of Health and Human Services reports that approximately $371 million is charged annually in connection with 15,436 hospital discharges for ankle OA. An increased understanding of ankle OA's effect on joint function will allow more informed assessment of gait and enhance treatment options to potentially slow disease progression.

METHODOLOGY

Ten patients with end-stage (surgical candidacy) ankle OA and ten matched controls were recruited within the guidelines of the Institutional Review Board. Informed consent was obtained from each subject prior to participation. Subjects were excluded if they had experienced a recent (<1 year) surgical, neurological, metabolic or lower limb musculoskeletal condition in the lower extremities, or had experienced multiple corrective surgeries. Subjects were asked to walk at a self-selected speed while barefoot over a 12 m walkway. A standard lower extremity marker set and 12-camera Vicon MX system were used to record lower limb segment position at 120Hz. Ground reaction force (GRF) data were recorded at 1200 Hz from 4 force platforms (Bertec) embedded in the walkway. Joint kinematics and kinetics were calculated using Plug-In Gait software (Vicon). Discrete values were extracted from kinematic and kinetic data specific to four phases of stance, based on Perry's definitions: load response (LR), midstance (MSt), terminal stance (TSt) and preswing (PSw). Angular position at initial contact (IC) and toe-off (TO) were also recorded. Group comparisons were made using an independent sample t-test. Within group comparisons were made using a paired student's t-test.

RESULTS

Only statistically significant differences ($p < 0.05$) are reported. Patients walked slower, with shorter stride length and reduced cadence compared to CON (Table 1). Ankle ROM in AFF was less than in UN for the OA group overall, and in all phases except TSt. Hip flexion was greater at IC in AFF; knee flexion and ankle plantarflexion were reduced at TO. CON subjects had greater knee ROM than AFF and greater ankle ROM than both limbs of the patient group. Ankle ROM in CON was greater than AFF in all phases except TSt. Hip flexion at IC was greater in CON than either AFF or UN, and knee flexion was greater at IC in CON compared to UN. At TO CON had greater hip flexion than AFF or UN and greater ankle plantarflexion than AFF. Compressive forces were decreased in all joints of AFF compared with UN and CON. The ankle dorsiflexor moment of AFF was reduced in LR compared to UN. The CON ankle plantarflexor moment in TSt/PSw was greater than in AFF. Knee extensor moment during MSt for CON was greater than AFF or UN, as was the hip extensor moment during LR. Ankle power generation in AFF was reduced in TSt/PSw compared to UN. Ankle power absorption during LR was greater in CON than in AFF or UN, as was power generation during TSt/PSw (Figure 1). Knee power absorption in CON during LR and PSw was greater than in AFF.

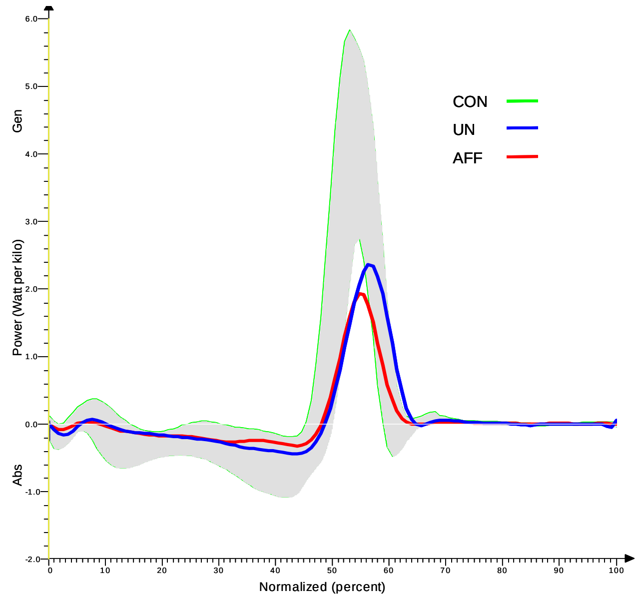


Figure 1. Ankle power group comparison.

Table 1. Subject demographics and TD parameters; mean (SD). Significant p-values are italicized.

	Age (yrs)	Height (m)	Mass (kg)	Gait Velocity (m/s)	Stride Length (m)	Cadence (steps/min)	Stance Time (% of cycle)
Patient (n=10)	50.9 (11.0)	1.7 (0.1)	94.5 (17.8)	1.04 (0.15)	1.20 (0.10)	103.3 (10.7)	63.6 (3.4)
Control (n=10)	55.4 (10.1)	1.7 (0.1)	86.9 (17.3)	1.28 (0.16)	1.33 (0.13)	115.2 (10.7)	67.4 (11.5)
p-value	0.353	0.810	0.346	*0.002*	*0.015*	*0.043*	0.170

DISCUSSION

Our hypotheses were supported by the findings that patients with ankle OA had reduced ROM, reduced TD magnitude and altered joint kinetics. Previous studies have reported similar findings in TD parameters and ankle kinematics/kinetics [4,5]; however, this is the first report of differences in the hip or knee of patients with ankle OA. Moreover, there appears to be a progressive reduction in the CON, UN, and AFF limbs for the parameters of ankle ROM, ankle power generation, and compressive force in all joints. As TD values from each limb's cycle were found to be equivalent, the progressive reduction cannot be explained by differences in walking speed between patient and control groups.

REFERENCES

[1] Huch, K. et al. Semin Arthritis Rheum 1997; 26:667-74.
[2] Baumhauer, J.F. et al. Am J Sports Med 1995; 23:564-70.
[3] Valtola, A. et al. Bone 2002; 30:238-42.
[4] Khazzam, M. et al. Gait Posture 2006; 24:85-93.
[5] Valderrabano, V. et al. Clin Biomech 2007; 22:894-904.

ACKNOWLEDGEMENTS

This study was supported by Department of Veterans Affairs grant A4513R.

Orthotic control of hyperpronation in the rearfoot is improved with exercise

David Hudson[1] and James Richards[2]

[1] Department of Physical Therapy, Western Carolina University, Cullowhee, NC
[2] Health and Exercise Sciences, University of Delaware, Newark, DE

Introduction: Plantar orthoses are well known to help control abnormal rearfoot motion, and have been shown to provide symptomatic relief in patients[1, 2]. Exercise has also been found to help control abnormal rearfoot motion[3]. It is reasonable to believe that the combined therapy of orthotics and exercise could be a superior treatment for problems related to hyperpronation. The purpose of this study was to determine the isolated and combined effect of orthotics and therapeutic exercises on controlling rearfoot hyperpronation. ***Clinical Significance:*** Abnormal pronation has been implicated in the etiology of numerous maladies of the lower extremities that can severely limit recreational, work, and routine daily activities of adults and children. Not all patients respond to orthotic therapy and many are in need of alternative treatments for pronation control due to high ‘out of pocket’ costs of custom orthoses. ***Methods:*** Six subjects (4 males and 2 females; age: 23.4 ±2.3; BMI: 25.1 ±2.0) with greater than 7° rearfoot pronation volunteered for the study and were prescribed custom, semi-rigid orthoses (Langer Inc., NY, USA). Subjects received their orthotics and were given a four-week break-in period, after which they returned for the first of two data collections. Subjects walked on a treadmill at a self-selected walking speed with and without orthoses while wearing their own running shoes. Subjects had six markers on right lower extremity (lateral knee, shank, lateral ankle, proximal and distal calcaneus and forefoot. The trajectory of the six markers were recorded by eight high-speed video cameras (Motion Analysis Corp., CA, USA) operating at 120 Hz. Rearfoot eversion excursion (RFγ) was calculated by custom software using Euler’s equations with the frontal plane motion calculated first. Following the collection of motion data, subjects were evaluated in order to set a baseline for a therapeutic program that included strengthening, stretching, and proprioception exercises for the lower extremity. Strengthening exercises targeted the lower extremity extensors, plantar flexors, hip abductors and hip external rotators; stretching exercise targeted the plantar flexors. Over the following four weeks subjects completed supervised therapy sessions three days/week. Exercise programs were progressed as the subjects were able to tolerate. Subjects were instructed to perform the stretching and proprioception exercises at home once per day on the days that they did not attend the supervised exercise sessions, and they were instructed not to change their current exercise programs during the four-week training period. After the four-weeks of exercise, subjects returned for the final data collection. Walking analyses were conducted as on the first trial; subjects walked at the same speed as on the baseline data collection. *Statistical Analysis:* Data were studied using a 2x2 factorial ANOVA with repeated measures on both factors: exercise treatment (baseline and post-exercise) and orthotic treatment (with orthoses and without orthoses). Statistical differences of $p \leq 0.05$ were considered significant.

***Results*:** Average RFγ values are summarized in **Table 1.** The 2x2 ANOVA found that both main effects were significant (exercise treatment $p = 0.018$ and orthotic treatment $p < 0.001$); the interaction (exercise x orthoses) was not significant.

Table1. Average rearfoot excursion (RFγ).

Condition	Baseline	Post-exercise
with Orthoses	11.0(2.5)	9.7(2.1)
without Orthoses	14.2(1.9)	12.8(1.7)

Discussion: These data provide evidence that suggests plantar orthoses help to reduce hyperpronation. Therapeutic exercise is most effective as an augmentation to orthotic treatment; however can be used independently to treat hyperpronation. Supervised exercise sessions were used in this study to ensure compliance; however, all of the exercises can be done at home without direct supervision. Studies involving more subjects and subjects with clinical symptoms are needed to determine the independent effects of the different modes of exercise (strengthening, stretching and proprioception), and if the changes found here are clinically significant.

References

1. Eng JJ, Pierrynowski MR. Evaluation of soft foot orthotics in the treatment of patellofemoral pain syndrome. *Phys Ther*. Feb 1993;73(2):62-68; discussion 68-70.

2. Pitman D, Jack D. A clinical investigation to determine the effectiveness of biomechanical foot orthoses as initial treatment for patellofemoral pain syndrome. *J Prosthet Orthot.* 2000;12:110-116.

3. Feltner ME, MacRae HS, MacRae PG, et al. Strength training effects on rearfoot motion in running. *Med Sci Sports Exerc.* Aug 1994;26(8):1021-1027.

Efficacy of an AFO during Gait in Adults with Hemiplegia

Karen J. Nolan [1,2], Mat Yarossi [1], Arvind Ramanujam [1], Krupa K. Savalia [1],
Gail F. Forrest [1,2], & Elie P. Elovic [1,2]
Kessler Medical Rehabilitation Research and Education Center, West Orange, NJ, USA[1]
UMDNJ - New Jersey Medical School, Newark, NJ, USA[2]
E-mail: knolan@kmrrec.org Web: www.kmrrec.org

Introduction
Hemiplegia is seen in fifty percent of stroke survivors and can cause disturbances in walking patterns due to impaired motor control, and sensation.[1] Spasticity, reduced motor control and/or ankle contracture often result in less dorsiflexion at footstrike and mid-swing. As a result, the affected side can strike the ground with the toes or side of the foot rather than the heel. [2, 3] During the swing phase this lack of dorsiflexion impairs the ability of the foot to clear the floor. [2, 4] Ankle foot orthotics (AFOs) are often prescribed to stroke survivors to improve functional ambulation (defined as improved walking speed, step length, cadence, etc). [5] The objective of this investigation was to quantitatively evaluate the functional and mechanical efficacy of a solid AFO during walking in individuals with hemiplegia secondary to stroke.

Clinical Significance
AFOs are often prescribed to individuals to assist with motor deficits and ambulation after stroke. Although AFOs have been linked to improved functional ambulation there is limited comprehensive data demonstrating the mechanical changes associated with AFO utilization. It is important to understand the relationship between mechanical and functional ambulation in order to fully understand the improvement in overall function.

Methods
Five participants with hemiplegia secondary to stroke (>1 year post) currently using an AFO during ambulation (at least 50% time) were recruited for participation. Subjects performed walking trials at a self selected pace, in two conditions (5 trials per condition): 1) with and 2) without the AFO. Kinematic data was collected at 120Hz (Vicon, Oxford Metrics, Oxford, UK) and the order of conditions was randomly assigned. Bilateral sagittal plane kinematics at the ankle, knee and hip, and temporal spatial gait parameters were used for analysis.

Table 1. Subject Characteristics

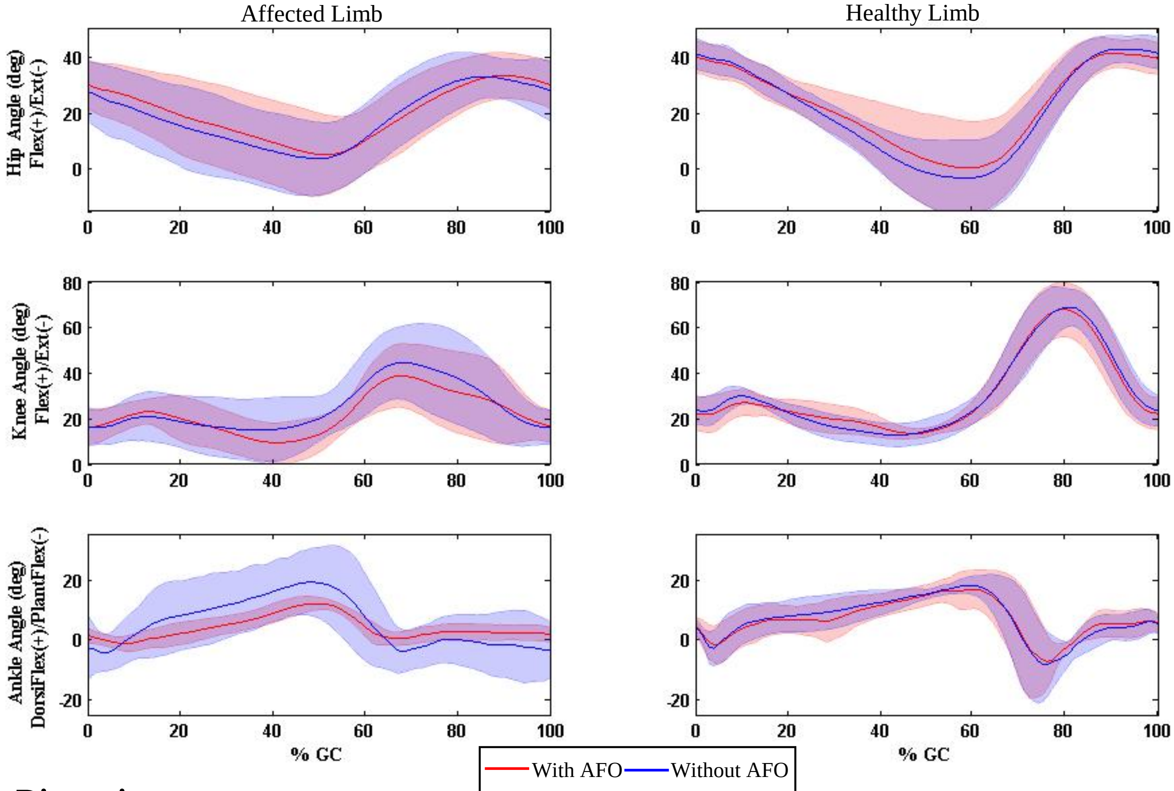
	Age (yrs.)	Gender	Height (in.)	Weight (kg)	LE Fugl-Meyer
Mean	52.80	4 M, 1 F	67.20	181.10	46.00
sd	± 12.68	-	± 3.35	± 47.37	± 8.12

Results
There were no significant changes in bilateral temporal spatial parameters with or without the AFO (Table 2). Bilateral hip range of motion (ROM) and healthy limb knee and ankle ROM remained unchanged with and without the AFO. Peak knee flexion angle on the affected side decreased (52.36 deg compared to 47.02 deg) with the AFO. At footstrike with the AFO, ankle dorsiflexion angle increased 9 deg, and throughout swing with the AFO planterflexion decreased 10 deg.

Table 2. Temporal Spatial Data

Difference (with AFO – without AFO)	IDS (%GC)	SS (%GC)	TDS (%GC)	SWING (%GC)	Stride Length (mm)	Step Length (mm)	Step Width (mm)	Foot Velocity (m/s)	Cadence (steps/min)
Affected	1.20	1.08	-2.85	0.57	43.50	17.83	-23.85	0.78	0.80
Healthy	-2.52	-0.20	1.66	1.06	32.54	17.50	-4.62	0.27	0.70

Figure 1. Sagittal Plane Kinematics



Discussion

The pilot data did not establish functional or mechanical efficacy for brace utilization. The AFO established dorsiflexion to provide foot clearance during swing and dorsiflexion at footstrike but the kinematic changes related to the AFO were restricted to the proximal segments. Specifically, the AFO altered kinematics at the ankle and knee without altering hip sagittal plane motion and therefore limited the improvements in temporal spatial parameters. Clinically, ankle ROM decreased throughout the gait cycle and knee flexion decreased during mid and terminal stance. The design of the AFO prevented foot drop by mechanically placing the ankle in a neutral position but functional ROM was restricted. Further research is needed to identify how an AFO affects individuals with chronic stroke.

References

1. Kelly-Hayes, M., et al (2003) J Stroke and Cerebrovascular Diseases.12:119-126.
2. Gok, H., et al (2003) Clin Rehabil. 17:137-139.
3. Lehmann, J. F., et al (1986) Arch Phys Med Rehabil. 67:380-386.
4. Mojica, J. A., et al (1988) Tohoku J Exp Med. 156:395-401.
5. Danielsson, A. et al (2004) J Rehabil Med. 36:165-168.

Acknowledgements: NIDRR Mary E. Switzer Fellowship Grant #H133F070028

GROUND REACTION FORCES DURING WALKING WITH TWO PROSTHETIC FEET: A PRELIMINARY STUDY

Alison J. Behrens[1], Jeremy D. Smith[1], Jerome R. Rifkin[2], and Gary D. Heise[1]
[1]The University of Northern Colorado, Greeley, CO contact: behrens.alison@gmail.com
[2]Transegrity Prosthetics, Inc., Boulder, CO

Introduction

Ground reaction force (GRF) magnitudes of the prosthetic leg have consistently been reported as reduced in comparison to those of the intact leg during walking [1]. In general, GRF profiles of unilateral, transtibial amputees suggest that the prosthetic leg bears less weight and generates less propulsive force prior to toe-off compared to the intact leg [1, 2]. To address the diminished propulsive force, a dynamic elastic response (DER) foot is commonly prescribed for individuals with an ability to modulate cadence and deal with obstacles. However, the carbon fiber structure of the DER-type foot limits the amount of rollover typically observed in an intact foot. The K3 Promoter prosthetic foot was designed to better mimic the natural roll over occurring in an intact foot by designing a forefoot articulation. This additional articulation is thought to lead to increased balance and improved ambulation on uneven terrain. The purpose of this study was to investigate the influence of the K3 Promoter foot on GRF patterns in a unilateral, transtibial amputee who currently uses a Ceterus® foot (i.e., DER).

Clinical significance

The K3 Promoter was designed to mimic the behavior of a fully intact foot. Walking with the K3 Promoter foot resulted in increased weight acceptance on the prosthetic leg, reduced braking forces on the intact leg, and unaltered propulsive forces for both legs. With further research and development, the K3 Promoter appears to have the potential for increasing the options available for prosthetic prescriptions.

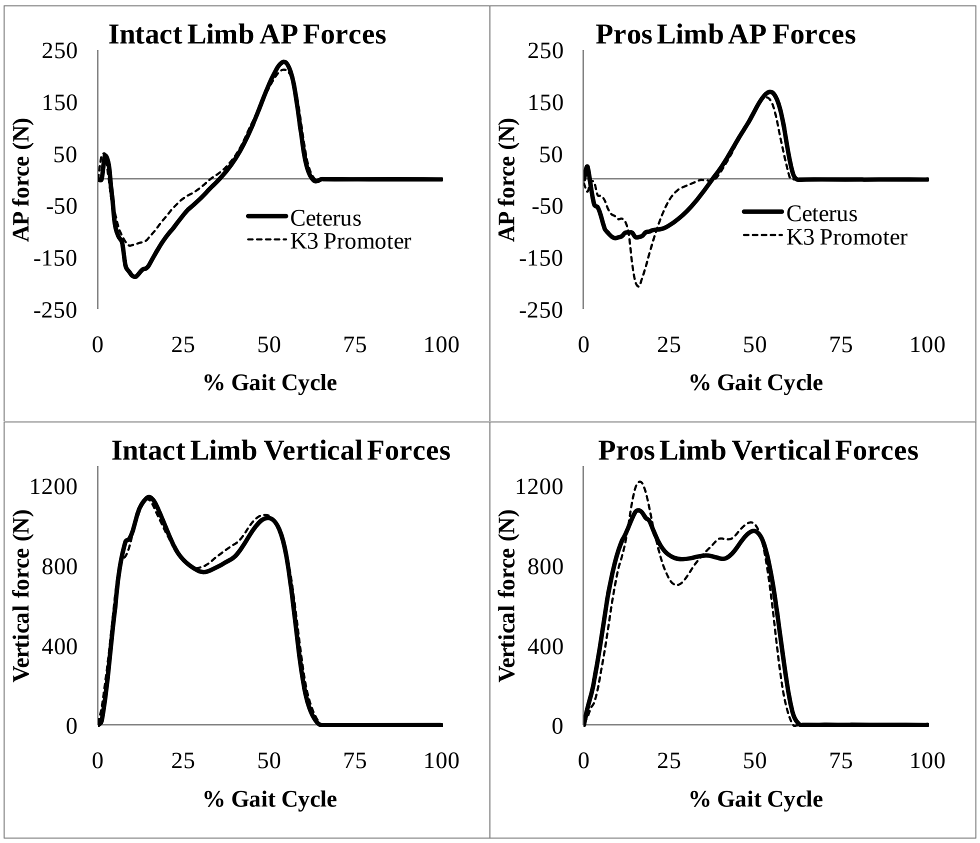
Methods

One male unilateral, transtibial amputee (age = 59 years, mass = 104.1 kg) participated in this study. The subject's amputation occurred in 2005 and was due to a post-operative infection at the ankle. From the time of his amputation the subject had been using a lock and pin suspension system with a Ceterus® prosthetic foot. The most recent prosthetic alignment occurred one week prior to testing for this study. The subject completed a series of overground walking trials at his preferred speed while wearing the Ceterus® and while wearing the K3 Promoter. GRF data were collected during walking trials. Vertical and anterior-posterior (AP) forces were averaged across two trials for purposes of data analysis.

Results

GRFs were different between prosthetic feet. In the AP direction, smaller peak (~32%) and average braking forces (~30%) were observed for the intact leg while walking with the K3 Promoter foot compared to the Ceterus® foot. For the prosthetic leg, walking with the K3 Promoter resulted in larger peak braking forces (~81%), but smaller average braking forces (~20%). Propulsive force differences in the AP direction were minimal. Vertical forces were similar in the intact leg for both foot conditions. However, in the prosthetic leg the first (~14%) and second (~5%) vertical force peaks were larger when walking with the K3 Promoter foot. Average vertical forces for the prosthetic limb, however, were less than 1% different.

Figure 1. GRFs of the intact and prosthetic limbs while walking with the Ceterus® and K3 Promoter prosthetic feet.



Discussion

Similar changes in GRFs have been reported following an increase in prosthetic foot mass up to 1.5 kg [3]. Thus, the greater mass (~1 kg) of the K3 Promoter foot likely contributed to our findings. However, given the magnitude change in the first vertical GRF peak (~145 N), it is unlikely that the greater foot mass fully explains the increased weight acceptance of the prosthetic leg. Reduced braking forces on the intact side suggest that forward momentum was maintained to a greater extent while walking with the K3 Promoter foot. Finally, although the K3 Promoter foot did not increase propulsive forces, they were similar in magnitude to those of the Ceterus® foot.

References

1. Nolan, L., et al. Gait Posture, 2003; 17(2): 142-151.
2. Winter, D.A. and S.E. Sienko J Biomech, 1988; 21(5): 361-367.
3. Hillery, S.C., et al. Prosthet Orthot Int, 1997; 21(2): 114-123.

The effects of wearing a functional limb length adjustment device with a short-leg walking boot on gait and plantar pressures: a pilot study

J. Song, B. Heilman, C. Kihm, C. Nimick, P. McDonald, and E. Zoltick
Temple University School of Podiatric Medicine

Introduction:

Short-leg walking boots are commonly used in treatment of injuries and post surgical management of the foot and ankle. Numerous studies show the benefits of short-leg walking boots (Pollo, 1999; Brown, 2004). The height of the boot's sole is typically greater than that of typical shoes, causing an artificial functional limb length discrepancy (LLD). Limb length discrepancies can cause adverse effects on gait, including limp, low back pain, and hip instability (Sathappan, 2007). Studies of short-leg walking boots reveal differences in gait mechanics, such as an elevated ground reaction force around midstance (Zhang, 2006; Shereff, 1985). Short-leg walking boots are worn for several months after surgery often causing the artificially induced LLD to alter gait in both limbs.

The *Evenup* (Even Up, Fort Myers, FL) is a device designed to attach to the contralateral shoe to compensate for the boot-induced functional LLD. The goal of the device is to produce a more normal gait by eliminating the functional LLD and avoiding the symptoms commonly associated with a LLD.

Clinical Significance:

The study objective was to determine if a contralateral shoe sole attachment eliminates the functional limb lengthening caused by wearing a short-leg boot.

Methods:

Ten healthy subjects were recruited for the study. Test procedures, benefits, and potential risk factors were explained and informed consent was obtained from each subject. Each subject underwent a biomechanical exam and was then tested in three conditions: (S) shoe-shoe, a pair of standard *New Balance 573* walking shoes, (B) shoe-boot, the same *New Balance* shoe on their left foot and a *Cozi™ Premium Walker* (DJ Orthopedics, Vista, CA) on their right foot, and (E) shoe+E-boot, the same *New Balance* shoe combined with the *Evenup* on their left foot and the same boot on their right foot.

In addition to the biomechanical exam, the following procedures were performed in the three testing conditions: bilateral measures of the functional limb length in the standing position from the ground to the ASIS, temporal and spatial footfall parameters were measured with the *Gaitmat II* system (E.Q., Inc, Chalfont, PA), and in-shoe plantar pressures were collected with the *Pedar-X* system (Novel, Munich, Germany).

Results:

The student's t-test was used with significance set at 0.05 to detect statistical significance between test condition pairs. The *Evenup* (-0.5 cm) caused significant differences in the functional LLD compared to the shoe-shoe condition (0.3 cm) and the shoe-boot condition (0.8 cm). The subjects' gait speed was statistically faster in the shoe-shoe condition (1.31 m/s) than the shoe-boot condition (1.18 m/s) and the shoe+E-boot condition (1.22 m/s). The pressure results can be seen in table 1 and the other temporal and footfall parameters are presented in table 2. Both boot conditions had similar plantar pressure measurements compared with the shoe-shoe condition. The pressures under the 5th metatarsal on the non-boot foot were significantly lower in both boot conditions, but this may result from the slower gait speeds. The pressure time integrals (PTI) under the boot foot were lower without the *Evenup*. The stance time and step length of the non-boot foot more closely matched the shoe-shoe condition while walking with the *Evenup*.

Table 1: Pressure Time Integrals (PTI) are separated by foot and shoe conditions. Statistical significance is indicated in gray.

PTI (N*sec/cm^2)		Ant./Post. Ratio		Med./Lat. Ratio		Sub-Hallucial		Sub-1st Metatarsal		Sub-5th Metatarsal	
Limb	Test Condition	mean	t-test	mean	t-test	mean	t-test	mean	t-test	mean	t-test
Boot Foot	Shoe-**Shoe** (S)	1.13	S-B	1.11	S-B	3.94	S-B	5.47	S-B	4.14	S-B
	Shoe-**Boot** (B)	1.07	S-E	0.98	S-E	2.38	S-E	3.60	S-E	3.49	S-E
	Shoe+E-**Boot** (E)	0.82	B-E	0.99	B-E	2.62	B-E	4.28	B-E	3.76	B-E
Non-Boot Foot	**Shoe**-Shoe (S)	1.22	no significance	1.11	no significance	4.68	no significance	5.02	no significance	4.34	S-B
	Shoe-Boot (B)	1.20		1.11		4.57		5.10		3.97	S-E
	Shoe+E-Boot (E)	1.15		1.14		4.40		4.92		3.66	B-E

Table 2: Temporal and spatial footfall results.

		Stance Time (sec)		Base of Support (m)		Step Length (m)	
Limb	Test Condition	mean	t-test	mean	t-test	mean	t-test
Boot Foot	Shoe-**Shoe** (S)	0.663	S-B	0.095	S-B	0.705	S-B
	Shoe-**Boot** (B)	0.691	S-E	0.121	S-E	0.669	S-E
	Shoe+E-**Boot** (E)	0.664	B-E	0.124	B-E	0.671	B-E
Non-Boot Foot	**Shoe**-Shoe (S)	0.667	S-B	0.072	S-B	0.697	S-B
	Shoe-Boot (B)	0.728	S-E	0.103	S-E	0.679	S-E
	Shoe+E-Boot (E)	0.714	B-E	0.105	B-E	0.711	B-E

Discussion:

The *Evenup* does create differences in walking when comparing the two boot conditions. The shoe's sole thickness was almost equal to the thickness of the boot, so adding the *Evenup* created an opposite LLD. The *Evenup* is height adjustable, so the height of the boot should be considered when optimizing the combined *Evenup* and shoe height. The *Evenup* is likely to affect joint kinematics/kinetics and alter gait beyond immediate effects, but those measures were not studied. The *Evenup* is a viable option for patients with limb lengthening sequelae.

References:

Pollo F, et al. *Orthopedics*. 1999, 22(5): 503-7.
Brown D, et al. *Arch Phys Med Rehabil*. 2004, 85: 81-6.
Sathappan S , et al. *J Arthroplasty*. 2008, 23(2): 203-209.
Zhang S, et al. *Gait and Posture*. 2006, 24: 487-492.
Shereff MJ, et al. *Clin Orthop Relat Res*. 1985, (192): 260-7.

Acknowledgements:

Even Up (Fort Myers, FL) for partial funding.

GAIT ANALYSIS POST HEMIPELVECTOMY/SACRECTOMY WITH SPINOPELVIC ARTHRODESIS RECONSTRUCTION

Krista A. ColemanWood, Michael J. Yaszemski, Franklin H. Sim, and Kenton R. Kaufman
[1]Division of Orthopedic Research

PATIENT HISTORY

At age 28 the patient was referred to our medical facility with a diagnosis of sacral and pelvic chondrosarcoma (grade II/III). Past patients with this diagnosis have had the lesion removed by hemipelvectomy and sacrectomy which, without successful reconstruction, left them wheelchair bound with only soft tissue connecting their vertebral column with their remaining limb.

CLINICAL DATA

The lesion was located over the central and left sacrum through the full thickness of the sacrum and extended into the fifth lumbar vertebral body and left iliac bone. An external hemipelvectomy and sacrectomy was performed on one day with reconstruction of a spinopelvic arthrodesis between the forth lumbar vertebra and the right hemipelvis five days later. (Figure 1) The spinopelvic arthrodesis was constructed from the previously removed proximal femur which was inverted and fitted into a notch prepared on the medial surface of the right iliac and attached with screw fixation. Additional support to the region was provided by a lumbar fusion with custom formed rods. The patient was immobilized with a hip spica cast and non-weight bearing for 3 months until osseous union of the spinopelvic arthrodesis was confirmed. The patient was fitted with a standard hemipelvectomy bucket prosthesis which, at 7kg, was very heavy compared to her body weight (43kg) and difficult for her to use.

Figure 1. Frontal view of 3D CT data rendering showing spinopelvic reconstruction.

GAIT DATA

The patient came to the Motion Analysis Laboratory 15 months post amputation to determine if walking without the prosthesis, using only axillary crutches, would cause excessive torque through the spinopelvic arthrodesis. Markers were placed using our standard modified Helen Hayes marker set with the following exceptions; the sacral marker was placed inferior of the central vertebral spinous processes at the level of the right PSIS, the left ASIS was placed at the same vertical level as the right ASIS on the residual tissue mass and all left thigh, shank and foot markers were placed on the left axillary crutch. Additional tracking markers were added to the posterior lateral pelvic region bilaterally. Data was acquired using a 10 camera Motion Analysis System (Motion Analysis Corporation, Santa Rosa, CA) as the patient walked with axillary crutches over three force platforms (2 Kistler [Kistler Instrument Corporation, Amherst, NY] and 1 AMTI [Advanced Mechanical Technology, Inc., Watertown, MA]) along a flat walkway and as the patient stood on a force platform with light touch contact for balance. Location of the spinopelvic arthrodesis with respect to the surface markers was estimated from CT image reconstruction measurements and a virtual marker was created in Visual 3D (C-Motion, Inc., Germantown, MD) to compare with the location of the projection of the ground reaction force vector during the quiet standing.

TREATMENT DECISIONS AND INDICATIONS

Shown in Figure 2, the linear extension of the ground reaction force vector during quiet standing demonstrates that forces are nearly coincident with the estimated location of the spinopelvic arthrodesis. Kinematics (Figure 3) during a three point swing through gait pattern show more adduction at the hip during late stance than laboratory norms. However, internal abduction moments (Figure 4) at the hip are very similar to laboratory norms. Moments in the sagittal and frontal planes are much lower than laboratory norms. (Figure 4)

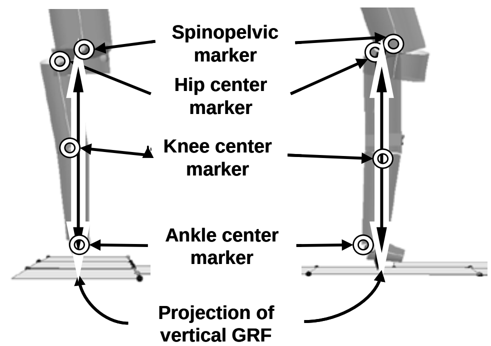


Figure 2. Frontal and sagittal plane projection of vertical ground reaction force with virtual marker location of spinopelvic arthrodesis.

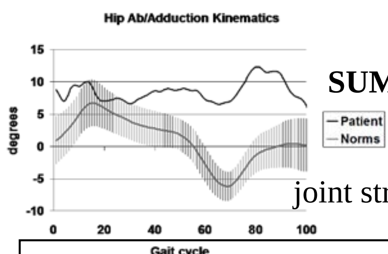


Figure 3. Frontal plane hip kinematics during three point swing through gait.

SUMMARY

Upright posture and weight bearing through the lower extremity joints is important for maintaining bone density and joint structures which are dependent on joint fluid movement for their nutrition. The motion study for this patient clearly demonstrated that forces pass through the spinopelvic arthrodesis in a manner which does not create excessive bending moments that would be detrimental to the reconstruction. Although kinematics demonstrated increased hip adduction, kinetics show abduction moments which are within the range of laboratory norms. It is likely that the redistribution of body mass caused by the amputation changed the loading at the hip so that even though the hip is in greater adduction, the demand on the hip abductor muscles is not increased. The sagittal and transverse plane moments are lower than those found in normal subjects. We conclude that this patient can safely walk using only axillary crutches without causing detrimental bending moments at the spinopelvic arthrodesis.

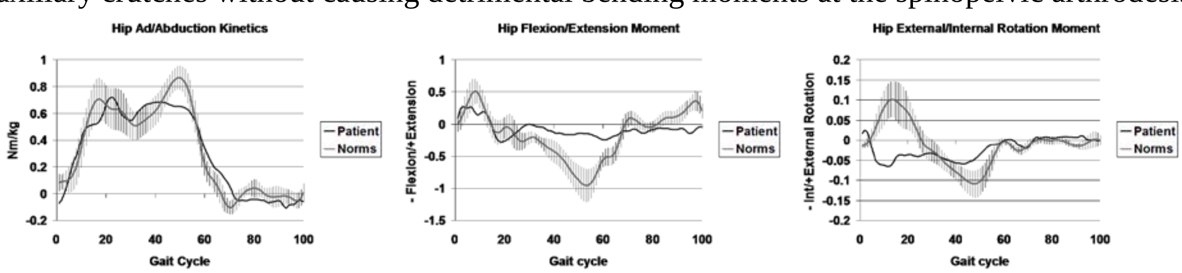


Figure 4. Frontal, sagittal and transverse plane hip kinetics during three point swing through gait

ACKNOWLEDGEMENTS

Special thanks to Diana K. Hanson, Alexander W. Hooke for their collaborative work.

Test-retest reliability of principal component scores for the knee adduction moment during gait in patients with knee osteoarthritis

Crystal Kean, Dr. Trevor B. Birmingham, Ian C. Jones, Dr. J. Robert Giffin
Wolf Orthopaedic Biomechanics Laboratory, University of Western Ontario

Introduction: The external knee adduction moment during gait is a proxy for medial compartment load, a risk factor for progression of osteoarthritis (OA) and an outcome measure for various treatment strategies(3, 7). Principal component analysis (PCA) is a statistical technique used to reduce the dimensionality of large data sets into a smaller set of meaningful component variables. PCA enables the use of the entire gait waveform, rather than selecting discrete points such as the peak, and may provide greater sensitivity to detect differences among patients or after interventions. However, PCA can be unstable (1) and the reliability of PCA when applied to gait waveform data is presently unclear. Therefore, the objective of the present study was to evaluate the test-retest reliability of principal component scores derived from the knee adduction moment during gait in patients with medial compartment knee OA.

Clinical Significance: Principal component scores for the knee adduction moment during gait appear to provide valuable measures for patients with knee OA. Reliability of these scores needs to be determined if they are to be used to evaluate treatments.

Methods: Twenty-four patients with medial compartment knee OA were tested on two occasions separated by at least 24 hours and within 1 week. During each test session, 3-dimensional gait analyses were completed using an 8-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA) synchronized with a single floor-mounted force plate (AMTI, Watertown, MA). Passive-reflective markers were placed on participants using a modified Helen Hayes marker set (5). Participants walked barefoot at their typical walking speed while kinetic and kinematic data were collected. Walking trials were completed until a total of five trials with clean force plate strikes were obtained.

The external knee adduction moment was calculated from the kinematic and kinetic data using commercial software (Orthotrak 6.0; Motion Analysis Corporation, Santa Rosa, CA) and custom post processing and data reduction techniques. The knee adduction moment waveform was normalized to time (0-100% gait cycle) and body size (% BW*Ht).

The normalized waveform data was organized into a matrix (n x p) where *n* was the total number of subjects and *p* indicated the gait cycle data points. PCA was then performed using MATLAB (Version7, MathWorks Inc.). In summary, PCA involved an eigenvector decomposition of the covariance matrix resulting in the principal components (PCs) that explained the waveform variance. PCs that together explained greater than 90% of variance were retained (4). Individual PC-scores for each PC were then calculated. The PC-scores provided a measure of how accurately the individual patient's waveform was projected onto the PC. For each PC retained, the test-retest reliability of patients' scores was evaluated using the intraclass correlation coefficient (ICC type 2, 1) (6).

Results: Three PCs were extracted from the original data set and together explained 94% of the waveform variance. PC1 explained 80% of the variance and captured the overall magnitude of the adduction moment throughout the gait cycle (Figure 1A). PC2 explained 8% of the variance and captured the first peak knee adduction moment occurring in initial period of stance (Figure 1B). PC3 explained 6% of the variance and captured the difference between early, mid and late stance (Figure 1C). The ICCs ranged from 0.78 to 0.86. PC1 had the highest ICC with 0.86 (95%CI: 0.71-0.94) while PC3 had the lowest ICC with 0.78 (95%CI: 0.55-0.90).

Discussion: The present findings suggest appropriate test-retest reliability of principal component scores for the knee adduction moment during gait in patients with medial compartment knee OA. Reliability was comparable to previously published values for the peak knee adduction moment (2).

References:

1. Al-Ibrahim, A.H., Al-Kandari, N.M. Stability of principal components. *Computational Statistics.* 2008; 23(1): 153-171.
2. Birmingham TB, Hunt MA, Jones IC, Jenkyn TR, Giffin JR. Test-retest reliability of the peak knee adduction moment during walking in patients with medial compartment knee osteoarthritis. *Arthritis Rheum.* 2007; 57(6): 1012-1017.
3. Hurwitz DE, Sumner DR, Andriacchi TP, Sugar DA. Dynamic knee loads during gait predict proximal tibial bone distribution. *J Biomech.* 1998; 31(5): 423-430.
4. Jackson JE. *A user's guide to principal components.* New York: John Wiley & Sons; 1991.
5. Kadaba MP, Ramakrishnan HK, Wootten ME. Measurement of lower extremity kinematics during level walking. *J Orthop Res.* 1990; 8(3): 383-392.
6. Shrout PE, Fleiss JL. Intraclass correlations: uses in assessing rater reliability. *Psychol Bull.* 1979; 86(2): 420-428.
7. Thorp LE, Wimmer MA, Block JA, Moisio KC, Shott S, Goker B, Sumner DR. Bone mineral density in the proximal tibia varies as a function of static alignment and knee adduction angular momentum in individuals with medial knee osteoarthritis. *Bone.* 2006; 39(5): 1116-1122.

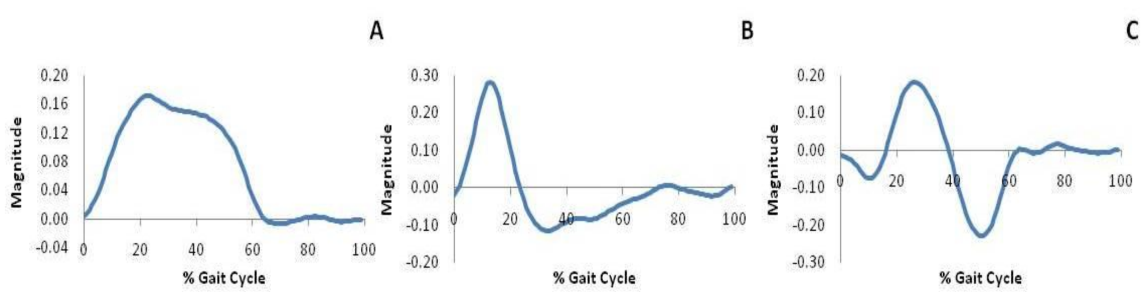


Figure 1: Retained PCA components. (A) PC1, (B) PC2 and (C) PC3.

Repeatability of Virtual Marker Based Foot Model in Adolescent Feet

Prabhav Saraswat[1,2], Bruce MacWilliams[1,2], Roy Davis[3]
[1]University of Utah, Shriners Hospitals for Children [2]Salt Lake City and [3]Greenville

Introduction:

Due to the close proximity of markers on the foot, marker placement variability is amplified in the angle calculations. A four-segment virtual anatomic marker based foot model addresses this problem by using small (4*mm* diameter) hemispherical markers to more accurately locate anatomic references for segment definition. Anatomic coordinate systems are determined relative to technical coordinate systems during a static trial, then the small anatomic markers are removed for the dynamic trials.

Clinical Significance:

Use of foot models in clinical decision-making is limited due to high measurement variability associated with them. A model with better repeatability can significantly improve the confidence in the results of such foot models.

Methods:

Nine pediatric subjects between the ages of 8 and 14 (average age 10.5 ± 1.61 years, 6 female, 3 male) were evaluated with two clinicians performing data collection on each subject. Five subjects were re-evaluated after four weeks to examine inter-session repeatability within clinician. A plaster cast of each subject's foot plantar surfaces was molded to hold the feet in the same position for every static trial[1, 2]. Small hemispherical markers (4*mm* D) were used to identify seven (2 on Shank, 1 on Hindfoot and 4 on Forefoot) key anatomic landmarks on each foot. A static trial was collected while subject kept their feet in the cast. This trial was used to record the position of the anatomic points relative to three reference markers fixed on the casting. This allowed the direct comparison of absolute segment attitude between sessions. Then a second static calibration trial was collected with nine technical markers (9*mm* D, 3 on each of the 3 segments of foot) and the seven anatomic markers on the foot. The position of anatomic landmarks was saved relative to the technical markers. A triad with three small (4*mm* D) markers was used to track the motion of hallux segment. No anatomic or virtual markers were used in hallux segment definition, only long axis of the triad was aligned with hallux during marker placement. The 4*mm* anatomic markers were removed and ten walking trials were collected.

Results:

The repeatability of the casting method was validated in an earlier study[2]. Figure 1 shows the intra-clinician and inter-clinician variability in absolute segmental attitude. Tibia rotation angle was calculated from anatomic markers on lateral and medial malleoli. Hindfoot axis definition used an anatomic marker on the calcaneal tubercle and goniometric measurement of standing ankle varus/valgus angle. Forefoot axis definition involved four anatomic markers: two along the midpoints of proximal and distal ends of second and third metatarsals and two along the most medial aspects of proximal and distal ends of first metatarsal. Both intra-clinician and inter-clinician average variability were less than 4^{o} for all angles, which is comparable to results reported in Davis *et al*.[2] in which a similar protocol was carried out on adult feet. As expected, inter-clinician variability were slightly larger than intra-clinician variability.

Figure 2 shows the average standard deviations (SD) of inter-segmental angles over a gait cycle for inter-trial, intra-clinician and inter-clinician variability. Variability is shown for three relative angles (hindfoot/shank, forefoot/hindfoot and hallux/forefoot). All segment definitions except hallux rely on anatomic markers and their average SD is within 6^o. The hallux segment attitude showed higher variability, which is expected because it was not defined with anatomic markers.

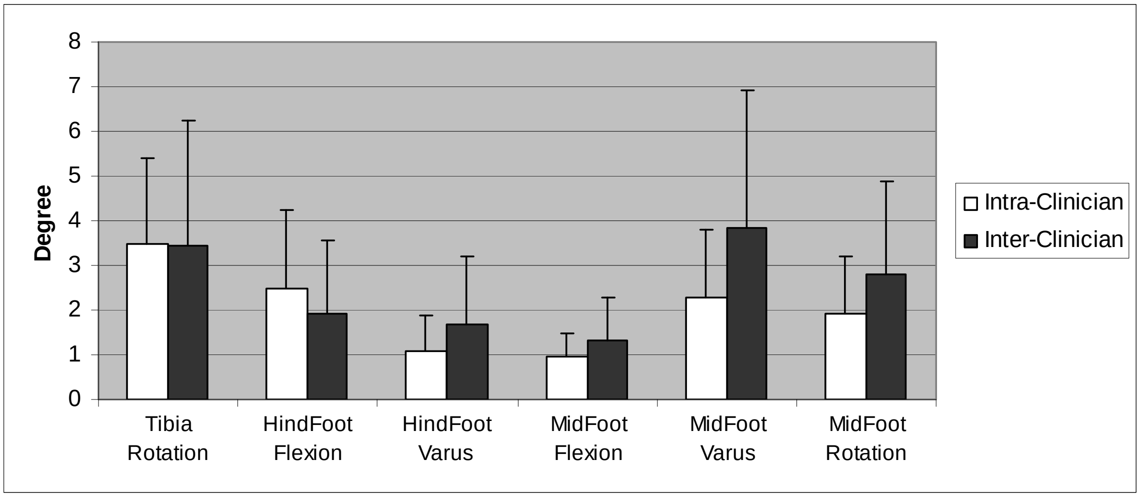


Figure 1: Variability in static segment attitude using virtual anatomic markers

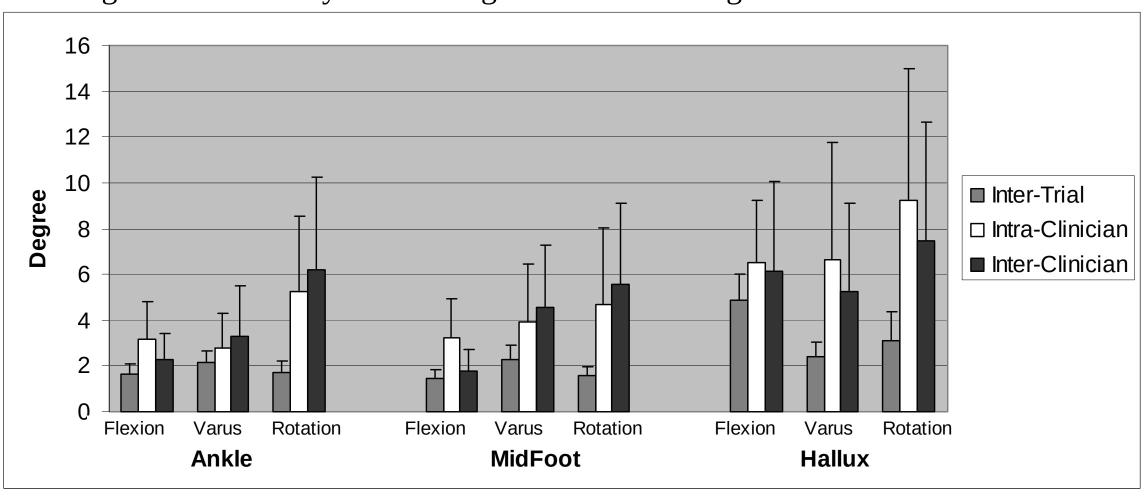


Figure 2: Average standard deviations over a gait cycle

Discussion:

Results show that the anatomic marker based model is as repeatable on adolescent feet as on adult feet. Small intra-clinician and inter-clinician variability in static segment definition resulted in acceptable standard deviations (2°-6°) of inter-segmental angles during gait in segments defined by markers placed on anatomic landmarks.

References:

1. Henley *et al.* (2008) Foot and Ankle Motion Analysis, CRC Press, pp. 445-464.

2. Davis *et al.* (2008), GCMAS Conference Proceedings, pp. 146-147.

Acknowledgements: The authors would like to acknowledge the time and assistance of Barbara Johnson, Soffe Lowell and Amy Shuckra. This work was supported by Shriners Hospitals for Children research grant #8954.

Functional Hip Joint Center calibration accuracy as compared with MRI images on 14 normal adults

Morgan Sangeux, Alana Peters, Richard Baker
Hugh Williamson Gait Laboratory, Royal Children's Hospital, Melbourne, Australia
Gait CCRE, Murdoch Childrens Research Institute, Melbourne, Australia
The school of Engineering, The University of Melbourne, Australia

INTRODUCTION

In gait analysis body segments of the patients are tracked thanks to skin mounted markers. The segments bones and joints positions are then deduced from these markers positions. The most common way to deduce the position of the hip joint centers (HJC) in clinical gait analysis is to use regression equations [1] that define the position of the hip from the patient anthropometric measurements and pelvis markers. A different approach uses joint functional calibration to locate joint centers in relation to the tracking markers. With this methodology, joint centers positions should mainly depend on segments relative movement and not on absolute marker placement. Although the accuracy of such methods have been previously reported [2, 3] there is still a need to know the HJC position accuracy in a clinical environment setting. This study assessed the functional HJC accuracy for different pelvis marker sets, functional movement amplitude and movement number of repetition on a normal adult population.

CLINICAL SIGNIFICANCE

Functional calibration to locate the HJC is seducing because it minimally relies on markers placement. In clinical gait analysis it means a better reliability of the data across assessors and labs which lead in turn into better comparison and share of normative data or case studies.

METHODS

14 adults (age: 36, range: 17–70 years, BMI: 23 range: 18–32 kg/m^2) participated to the study.

For the functional calibration of the hip joints the subjects were equipped with 4 markers on the pelvis (Left/Right and Anterior/Posterior ASIS) and 3 on each thigh, placed on the middle of the segment anteriorly and laterally. To evaluate the influence of the soft tissue artifact (STA) at the pelvis on the computation, HJC have been determined with a 4 ASIS markers pelvis model and a 3 markers model composed of the 2 PSIS and the opposite to the calibrated HJC side ASIS.

Segments kinematics was determined by the least square mapping [4] of the skin mounted markers and HJC position was determined using the method described in [3]. This method, recently reviewed and called CTT in [5] is one of the less sensitive to simulated STA.

The hip calibration exercise consists in 5 movements: hip flexion, hip extension, hip abduction, hip flexion and abduction and hip extension and abduction. To test the influence of the number of repetition the 5 movements were repeated 1 or 2 times. To assess the influence of the movement amplitude on the results they were performed at natural or restricted amplitude.

Subjects of both groups had an MRI (3T Siemens Trio, Germany) scan of their pelvis. The MRI captured volume included the anterior and posterior ASIS as well as the soft tissues on top of them. This enables to build a pelvis frame similar to the gait analysis one.

RESULTS

Difference between MRI and skin markers based ASIS width, PSIS width and pelvic depth were **31**(18), **9**(7) and **16**(8) mm respectively.

The following table gives the results on ΔHJC, the difference between MRI and functional inter HJC distance and the distance between MRI and functional HJC position [**Mean** (SD)].

Factor		ΔHJC (mm)		Diff HJC position (mm)	
Amplitude	Full	**10.3**	(4.3)	**27.9**	(9.7)
	Restricted	**11.4**	(6.7)	**35.3**	(10.3)
Repetition	1	**10.5**	(6.0)	**31.6**	(10.1)
	2	**11.2**	(6.1)	**29.6**	(9.7)
Model	4 pelvis markers	**9.7**	(5.9)	**38.6**	(11.6)
	3 pelvis markers	**12.0**	(6.7)	**24.8**	(8.9)

The values for the regression based estimation were **33**(16) mm for ΔHJC and **31**(6) mm for Diff HJC position.

DISCUSSION

In overall, ΔHJC results show that functional estimation is 3 times closer to the MRI measure than regression equation estimation. The different influencing factors studied do not alter the functional estimation of more than 2.3mm.

The distance between functional calibration HJC position and MRI based HJC position did not give a better estimation than the regression equation with an average 3cm difference. These unexpected values are greater than previously reported. Some influencing factors do play a role for this result with full amplitude movement being better than restricted one and 3 pelvis markers closer by 1.4cm than its 4 pelvis markers counterpart.

The comparison of MRI and skin markers based anthropometric measures exhibits large differences with a maximum of more than 3cm on the ASIS width measure. These differences could explain the contradiction of having a good agreement on ΔHJC, a measure not related to pelvis frame definition, and Diff HJC position a measure that directly depends on the similarity of MRI and marker based pelvis frame. The differences between MRI lying down and marker based standing up positions could explain this and on-going work are looking at this particular issue.

CONCLUSION

The good agreement of MRI and functional calibration ΔHJC results are promising. However the differences observed on the HJC positions as well as the pelvis frame dimensions needs to be thoroughly investigated.

REFERENCES

[1] Davis et al., 1991, Human Movement Science (10), 575-587

[2] Leardini et al., 1999, Journal of Biomechanics (32), 99-103

[3] Piazza et al., 2004, Journal of Biomechanics (37), 349-356

[4] Soderkvist et al., 1993, Journal of Biomechanics (26), 1473-1477

[5] Ehrig et al., 2006, Journal of Biomechanics (39), 2798-28

The Effect of Medial Markers on Knee Kinematics Measurements from Plug-in-Gait

Wong, P[1]; Callewaert, B[2]; Labey, L[1]; Leardini, A[3]; Desloovere, K[4]

[1]*European Centre for Knee Research, Smith & Nephew, Leuven, Belgium;* [2]*University Hospital Pellenberg, Katholieke Universiteit Leuven, Belgium;* [3]*Movement Analysis Laboratory, Istituto Ortopedico Rizzoli, Bologna, Italy;* [4]*Department of Rehabilitation Sciences, Katholieke Universiteit Leuven, Belgium;*

Introduction: It was hypothesized that additional markers on the medial aspects of the knee and ankle can improve the tracking of the underlying bones in standard gait analysis. As long as these markers are visible during dynamic trials, they are expected to allow more direct calculation of knee and ankle rotations out of the sagittal planes. A previous study reports reduced errors when these are used in several protocols and marker sets[1]. The present study compares the knee kinematics curves during gait resulting from two methods: (1) using a standard model and marker set (Plug-in-Gait[2] [PiG], Vicon®, Oxford, UK) with a knee alignment device[3] (KAD), and (2) using a PiG-based model with additional medial markers (MM). When PiG is used without the KAD, errors in knee joint rotations increase because of both knee axis malalignment and large skin motion artifacts at the thigh. The KAD was introduced to better define rotation axes[3], which reduces crosstalk error from axis malalignment and slightly improves these measurements. It does this by placing a constant angle offset on the axis throughout a dynamic trial, based on a static trial, which accounts for marker misplacement on the thigh and shank. The PiG-MM method used here is similar but applies a variable angle offset correction, instead of a constant offset, throughout an entire dynamic trial, which may better address crosstalk error. Additionally, the PiG-MM method may reduce error due to skin motion artifacts, when compared to the thigh and shank markers, by improving bone tracking. One criterion for judging the accuracy of the two methods is the range of motion (ROM) of ab/adduction measured during swing phase of normal walking; minimal variation in the coronal plane is expected in a nonpathologic knee. At the same time, higher repeatability also is desired in the measurements. Better methods ideally should improve both accuracy and repeatability.

Clinical Significance: This study compares methods for measuring *in vivo* knee kinematics noninvasively, more accurately and repeatably. The results have implications on how to better evaluate the knee before and after surgical interventions in patients.

Methods: Four healthy male volunteers (mean age = 35 ± 10 years, body mass = 82 ± 30 kg) underwent gait analysis sessions on different days, managed by a well-trained physical therapist, using the same marker configurations[2,3] including medial markers, with an 8-camera Vicon system and 2 AMTI forceplates. Subjects performed 3 walking trials per session over 3 sessions. Clinical knee rotations were calculated using the PiG-KAD[2] and PiG-MM methods. The MM method calculated the knee joint center as the midpoint of the medial and lateral markers, then found the plane normal to the femoral mechanical axis. The knee flexion axis was defined by projecting onto this plane the line joining the medial and lateral marker. This procedure was repeated for each frame of the motion trial, and an analogous process was applied to the ankle. The MM calculations were made by modifying the open-source "Golem" model (Vicon BodyBuilder) based on PiG. Data were normalized to a 100% gait cycle and averaged at each %cycle across all trials. Repeatability was estimated as the standard deviation of the errors from the averaged trial at each %cycle.

Results: 29 out of 36 gait trials collected were used in this study, as they had full marker visibility and no data collection anomalies. Figure 1 shows the average knee kinematics curves. The shapes of ab/adduction and in/external rotation curves differed slightly between the KAD and MM methods, with smoother curves resulting for the latter method. This method also offset the in/external rotation by about 20°, but this was expected since it aligned the ankle flexion axis according to the line through the malleoli, while PiG-KAD did not[3]. This offset is easily corrected but was not done here. Table 1 compares the repeatability of the two methods, as well as the resulting ranges of the average curves across the gait cycle.

Discussion: Regarding accuracy, the PiG-MM method shows smaller ROM in the coronal plane during swing phase and, therefore, may give a reasonably better estimate of knee kinematics. However, regarding repeatability the PiG-MM method produces larger variability in knee flexion, likely due to skin motion artifact at deeper flexion. Ultimately better tracking of the underlying bone will lead to better intra-subject accuracy and repeatability, and markers in addition to the medial marker may realize this. These results also suggest that different methods may be better for different research purposes. For the knee, the PiG-MM method may be used when sufficient cameras are available, no KAD is available, or when minimal motion in the coronal plane is expected.

Table 1. Inter-subject repeatability and ranges of kinematics.

		SD of Errors (°)	Range of average trial (°)
Flexion	**KAD**	4.8	62.5
	MM	8.1	62.0
Adduction	**KAD**	3.8	6.4
	MM	3.1	3.7
Rotation	**KAD**	4.6	11.1
	MM	4.2	10.3

References

1. Biagi F, et al. European Society of Movement Analysis Conference, 2008.
2. Kadaba MP, et al. *Journal of Orthopaedic Research*. 1990;8:383-92.
3. Schache AG, et al. *Gait & Posture*. 2006;24:100-109.

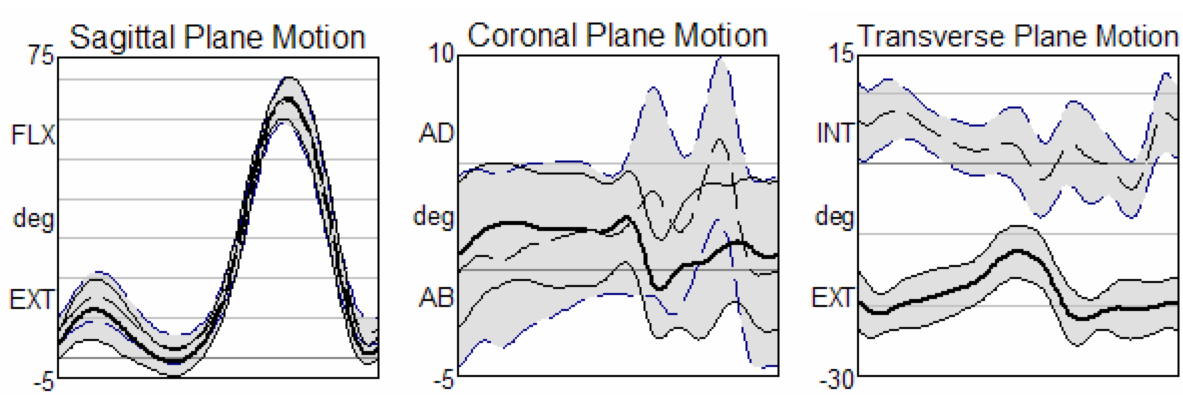


Figure 1. Average knee kinematics curves for the KAD (dashed line) and MM (solid line) methods (mean ± SD) across 100% gait cycle.

A Comparison of Two Functional Methods for Calculating Joint Centers and Axes in a Clinical Setting

Adam Rozumalski, MS [1,2] and Michael H. Schwartz, PhD [1,2]
[1]Gillette Children's Specialty Healthcare, St Paul, [2]University of Minnesota, Minneapolis, USA

INTRODUCTION

Functional model calibration – the process of using subject movements to estimate joint centers and effective axes of rotation – has been shown to be practical, objective, and reliable in the clinical setting [1]. A functional model calibration algorithm developed by Schwartz and Rozumalski (Schwartz Transformation Technique – or – STT) has been shown to be objective and repeatable [2]. However, the calculations can be time consuming – taking up to 6 minutes per subject using an optimized algorithm, and potentially much longer than that. The symmetrical center of rotation estimation (SCoRE) and symmetrical axis of rotation approach (SARA) are new functional model calibration techniques shown to be nearly identical to the STT when applied to simulated data and simulated marker noise [3, 4]. Both the SCoRE/SARA and STT algorithms are derived from the constraint that joint centers/axes must remain constant relative to the adjacent segments The present study compares the two methods in a clinical setting on actual patient data where the motions are not ideal, and the marker errors due to soft tissue artifact and reconstruction error are unknown, and likely to be complex.

CLINICAL SIGNIFICANCE

The SCoRE/SARA and STT methods have been shown to be equivalent when using simulated data. The SCoRE/SARA method has yet to be evaluated in the clinical setting.

METHODS

We retrospectively analyzed 1336 hip range-of-motion (ROM) trials and 1260 knee ROM trials for 647 patients seen for analysis in our gait analysis laboratory between January 1st, 2007 and September 1st, 2008. Approximately 75% of the patients had been diagnosed with cerebral palsy, while the other 25% had widely varying diagnoses. The SCoRE and STT methods were used to calculate the hip joint center, while the SARA and STT methods were used to calculate the knee joint axis. The distance between the hip centers from the two methods was compared, as well as the angle between the knee axes. Computation time was also examined.

RESULTS

The mean distance between the STT and SCoRE hip centers was 2.5 (3.0) mm, with no meaningful bias in any direction. The angle between the STT and SARA knee axes was 0.8° (1.8°) [Figure 1]. The SCoRE method was 212 times faster for computing hip centers and the SARA method was 218 times faster for computing knee axes.

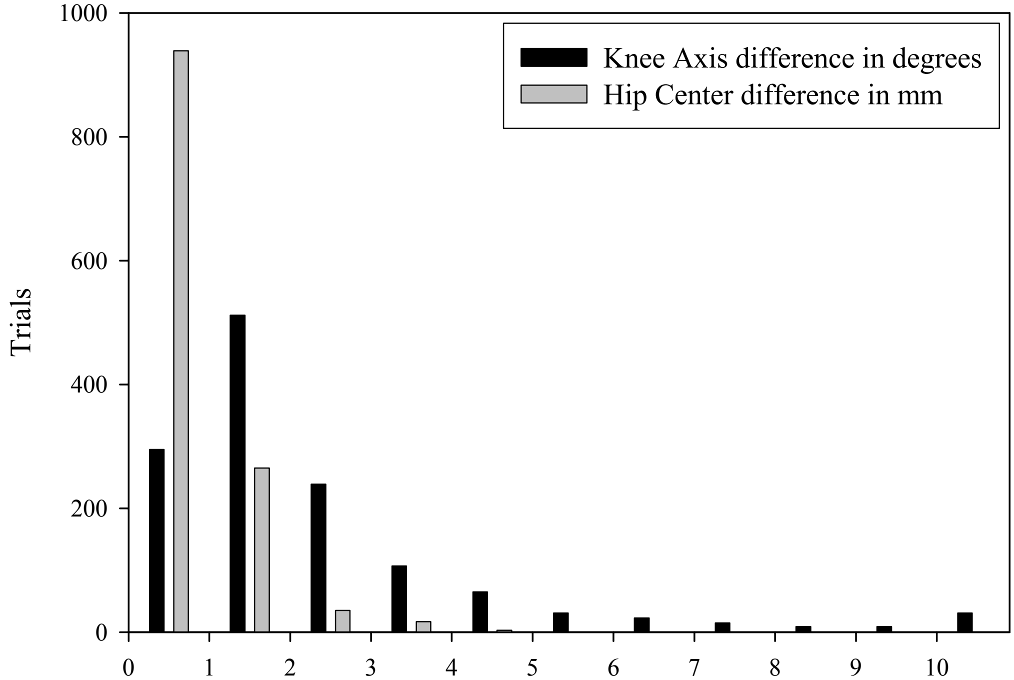


Figure 1. Comparison of STT and SCoRE/SARA methods. The x-axis is the difference between the hip centers (mm) and the knee axes (degrees) for the two methods. Overall, the two methods are seen to be practically equivalent.

DISCUSSION

Two functional model calibration methods, one developed by Schwartz and Rozumalski, and the other by Ehrig *et al*., produced nearly identical results using clinical data. This confirmed Ehrig's findings, which were based on simulated data and simulated noise. The subjects studied had a wide range of pathologies, body types, and joint ranges of motion, rendering this result quite robust. The SCoRE/SARA method was found to be over $200\times$ faster, and therefore seems preferable.

REFERENCES

1. Rozumalski A, Schwartz MH, 13th annual meeting of ESMAC 2004, Warsaw, Poland.
2. Schwartz MH, Rozumalski A, J Biomech. 2005 Jan;38(1):107-16.
3. Ehrig RM, *et al*., J Biomech. 2006;39(15):2798-809.
4. Ehrig RM, *et al*., J Biomech. 2007;40(10):2150-7.

COMPARISION OF TECHNIQUES FOR FINDING THE KNEE JOINT CENTER

Diane M. Serfling, Alexander W. Hooke, Kathie A. Bernhardt, Kenton R. Kaufman

Division of Orthopedic Research, Mayo Clinic, Rochester, MN, USA
E-mail: kaufman.kenton@mayo.edu

INTRODUCTION

The accuracy of gait studies is largely dependent on placement of the markers used [1, 2, 3]. Misplacement of markers can affect kinematics and kinetics, which in turn can affect clinical decision making. The goal of this study was to look at inter- and intra-relater reliability with traditional marker placement at the knee using manual palpation versus using a Knee Center Device (KCD) developed by Otto Bock Healthcare LP (Otto Bock Healthcare, Minneapolis, MN) [3].

CLINICAL SIGNIFICANCE

Consistency of knee marker placement is a primary factor in accurate gait analysis.

METHODS

Four physical therapists placed markers on a single subject that was free of knee pathologies. Experience in marker placement between therapists varied greatly (8 years, 6 years, 1.5 years, and 2 months). A modified Helen Hayes marker set was used [4]. All markers used for the gait analysis were left on the patient for all trials, with no variation or marker movement except for the lateral and medial knee markers. In random order, each therapist individually placed the knee markers on the subject's right knee using manual palpation[5] or the KCD. Data was collected as 4 sets of 3 trials, yielding 6 total trials for each condition, with the order of sets randomized across therapists. The KCD was used according to its established protocol and the knee centers identified with the device were then used for marker placement [3].

Objective gait measurements were acquired with a computerized video motion analysis system utilizing ten infrared cameras (EvaRT 5.0, Motion Analysis Corporation, Santa Rosa, CA). Reduction of data was done using Visual3D (C-Motion, Inc., Rockville, MD). Inter- and intra-therapist variability in lower extremity kinematics and kinetics were calculated at every point in the normalized gait cycle for both the KCD and palpation technique using the method proposed by Schwarz et al. [6]. A one-sample t-test was used to test whether the difference between palpation and KCD kinematics' standard deviation (σ_{KCD}-σ_{Palp}) differed significantly from zero. Statistical significance was set at ≤ 0.05.

RESULTS

Inter-Therapist	Ankle: PDF (°)	Hip: Flex/Ext (°)	Hip: Rotation (°)*	Knee: Var/Val (°)*	Knee: Flex/Ext (°)*	Knee: Rotation (°)*	Knee: Flex/Ext Moment (Nm/kg)*
Maximum	0.402	0.559	2.134	1.247	0.955	1.952	0.028
Mean	-0.020	0.020	1.085	0.288	0.121	1.160	0.004
Variance	0.027	0.040	0.426	0.129	0.116	0.157	0.000

Table 1: The maximum, mean, and variance of standard deviation differences ($\sigma_{\text{Knee Device}} - \sigma_{\text{Palp}}$), of lower extremity kinetics and kinematics. PDF = Planter/Dorsiflexion, Flex/Ext = Flexion/Extension, Var/Val = Varus/Valgus. * = difference significant at $p = 0.05$ level.

Significant inter-therapist differences were identified for the variables of hip rotation, knee varus/valgus, flexion/extension, and rotation angles, and knee flexion extension moment. For all these variables, variability using the KCD was greater than using manual palpation, indicating that device in less reliable than palpation between therapists.

Intra-therapist differences between the two techniques showed a total of 3 conditions where manual palpation had more variability than the KCD. In each of these cases, the difference was 0.01° which is not clinically significant. This indicates that the device is less, or equally reliable, within therapists as well.

SUMMARY/CONCLUSIONS

The Knee Center Device was less reliable between therapists than manual palpation. Therapists had limited experience in using the device prior to the study; however within-therapist analysis indicates that therapists were no more reliable with the device at an individual level and thus this is not the source of the between-therapist findings. As the KCD determines both the medical and lateral knee marker locations, it is possible that shifts in both locations influenced the results. While extensive training with the device may improve its reliability, our results indicate that these improvements would be negligible and not worth the cost such an endeavor would require.

REFERENCES

1. Gorton, G., et al. (2001) Gait and Posture, 13: 247.
2. Gorton, G., et al (2002) (Suppl 1) Gait and Posture, 16: S65-S66.
3. Blumentritt, S. (1998) Orthopidic-Technik, Dec: 938-945.
4. Davis, R. B. et al (1991). Human Movement Science, 10: 575-587.
5. Kabada,M.P., et al (1989) Journal of Orthopaedic Research 7: 849-860
6. Schwartz, M.H., et al. (2004) Gait and Posture, 20:196-203

ACKNOWLEGMENTS

Special thanks to Otto Bock Healthcare for the use of the Knee Center Device and the Mayo Foundation for support.

Functional Hip Joint Center calibration accuracy as compared with MRI images on 14 normal adults

Morgan Sangeux, Alana Peters, Richard Baker
Hugh Williamson Gait Laboratory, Royal Children's Hospital, Melbourne, Australia
Gait CCRE, Murdoch Childrens Research Institute, Melbourne, Australia
The school of Engineering, The University of Melbourne, Australia

INTRODUCTION

In gait analysis body segments of the patients are tracked thanks to skin mounted markers. The segments bones and joints positions are then deduced from these markers positions. The most common way to deduce the position of the hip joint centers (HJC) in clinical gait analysis is to use regression equations [1] that define the position of the hip from the patient anthropometric measurements and pelvis markers. A different approach uses joint functional calibration to locate joint centers in relation to the tracking markers. With this methodology, joint centers positions should mainly depend on segments relative movement and not on absolute marker placement. Although the accuracy of such methods have been previously reported [2, 3] there is still a need to know the HJC position accuracy in a clinical environment setting. This study assessed the functional HJC accuracy for different pelvis marker sets, functional movement amplitude and movement number of repetition on a normal adult population.

CLINICAL SIGNIFICANCE

Functional calibration to locate the HJC is seducing because it minimally relies on markers placement. In clinical gait analysis it means a better reliability of the data across assessors and labs which lead in turn into better comparison and share of normative data or case studies.

METHODS

14 adults (age: 36, range: 17–70 years, BMI: 23 range: 18–32 kg/m^2) participated to the study.

For the functional calibration of the hip joints the subjects were equipped with 4 markers on the pelvis (Left/Right and Anterior/Posterior ASIS) and 3 on each thigh, placed on the middle of the segment anteriorly and laterally. To evaluate the influence of the soft tissue artifact (STA) at the pelvis on the computation, HJC have been determined with a 4 ASIS markers pelvis model and a 3 markers model composed of the 2 PSIS and the opposite to the calibrated HJC side ASIS.

Segments kinematics was determined by the least square mapping [4] of the skin mounted markers and HJC position was determined using the method described in [3]. This method, recently reviewed and called CTT in [5] is one of the less sensitive to simulated STA.

The hip calibration exercise consists in 5 movements: hip flexion, hip extension, hip abduction, hip flexion and abduction and hip extension and abduction. To test the influence of the number of repetition the 5 movements were repeated 1 or 2 times. To assess the influence of the movement amplitude on the results they were performed at natural or restricted amplitude.

Subjects of both groups had an MRI (3T Siemens Trio, Germany) scan of their pelvis. The MRI captured volume included the anterior and posterior ASIS as well as the soft tissues on top of them. This enables to build a pelvis frame similar to the gait analysis one.

RESULTS

Difference between MRI and skin markers based ASIS width, PSIS width and pelvic depth were **31**(18), **9**(7) and **16**(8) mm respectively.

The following table gives the results on ΔHJC, the difference between MRI and functional inter HJC distance and the distance between MRI and functional HJC position [**Mean** (SD)].

Factor		ΔHJC (mm)		Diff HJC position (mm)	
Amplitude	Full	**10.3**	(4.3)	**27.9**	(9.7)
	Restricted	**11.4**	(6.7)	**35.3**	(10.3)
Repetition	1	**10.5**	(6.0)	**31.6**	(10.1)
	2	**11.2**	(6.1)	**29.6**	(9.7)
Model	4 pelvis markers	**9.7**	(5.9)	**38.6**	(11.6)
	3 pelvis markers	**12.0**	(6.7)	**24.8**	(8.9)

The values for the regression based estimation were **33**(16) mm for ΔHJC and **31**(6) mm for Diff HJC position.

DISCUSSION

In overall, ΔHJC results show that functional estimation is 3 times closer to the MRI measure than regression equation estimation. The different influencing factors studied do not alter the functional estimation of more than 2.3mm.

The distance between functional calibration HJC position and MRI based HJC position did not give a better estimation than the regression equation with an average 3cm difference. These unexpected values are greater than previously reported. Some influencing factors do play a role for this result with full amplitude movement being better than restricted one and 3 pelvis markers closer by 1.4cm than its 4 pelvis markers counterpart.

The comparison of MRI and skin markers based anthropometric measures exhibits large differences with a maximum of more than 3cm on the ASIS width measure. These differences could explain the contradiction of having a good agreement on ΔHJC, a measure not related to pelvis frame definition, and Diff HJC position a measure that directly depends on the similarity of MRI and marker based pelvis frame. The differences between MRI lying down and marker based standing up positions could explain this and on-going work are looking at this particular issue.

CONCLUSION

The good agreement of MRI and functional calibration ΔHJC results are promising. However the differences observed on the HJC positions as well as the pelvis frame dimensions needs to be thoroughly investigated.

REFERENCES

[1] Davis et al., 1991, Human Movement Science (10), 575-587

[2] Leardini et al., 1999, Journal of Biomechanics (32), 99-103

[3] Piazza et al., 2004, Journal of Biomechanics (37), 349-356

[4] Soderkvist et al., 1993, Journal of Biomechanics (26), 1473-1477

[5] Ehrig et al., 2006, Journal of Biomechanics (39), 2798-28

Mechanical energy generation, absorption, and transfer during walking in children

B.R. Umberger[1], S. Augsburger[2], J. Resig[2], D. Oeffinger[2], R. Shapiro[3], & C. Tylkowski[2]
[1]*Dept. of Kinesiology, Univ. of Massachusetts, Amherst, MA, USA*; [2]*Motion Analysis Laboratory, Shriners Hospital for Children, Lexington, KY, USA*; [3]*Dept. of Kinesiology & Health Promotion, Univ. of Kentucky, Lexington, KY, USA*

Introduction

During walking, the net moments acting about the joints generate mechanical energy, absorb mechanical energy, and cause mechanical energy to be transferred between body segments. Not only can a joint moment generate energy to (or absorb energy from) a distant body segment, but the magnitude of energy transfer can exceed the magnitude of energy generation/absorption [1]. Understanding the nature of these energy flows in normal walking is a critical first step towards a better appreciation of the causes of gait pathology. Fregly and Zajac [1] described an induced power technique that reveals how any force or moment affects the mechanical power of every segment in a model of seated pedaling. Their technique has subsequently been adapted and applied to the study of walking in healthy adults [2, 4]. In the current investigation, we used a similar approach to determine how the hip, knee, and ankle joint moments generate, absorb, and transfer mechanical energy during walking in able-bodied children.

Statement of clinical significance

The present induced power data obtained from able-bodied children will form the basis for future comparison with data collected on children who exhibit gait disorders.

Methods

Kinematic (60 Hz) and kinetic (960 Hz) data were collected on 33 able-bodied children (5-17 yr) as they walked along a 10 m walkway at their self-selected speed. The subjects were subdivided into young (5-7 yr, n=10), middle (8-11 yr, n=12), and older (12-17 yr, n=11) groups. Experimental data were processed using EvaRT 5.0 and OrthoTrak 6.3.4. These data served as inputs to a simulation model [5] that was used to perform the induced power analyses. The model consisted of seven rigid segments representing the right and left foot, shank, and thigh, as well as a lumped head-arms-trunk (HAT) segment. During periods of ground contact, the feet were constrained to the ground at the center of pressure. The model was actuated by joint moments at the hip (flexion/extension and abduction/adduction), knee (flexion/extension), and ankle (dorsi/plantarflexion). The model was scaled to the size of each subject, and then configured at the start of the gait cycle. The experimental moments were input to the model one at a time, and the mechanical power of each segment was computed. This process was repeated for each sample over the entire gait cycle. The thigh, shank, and foot powers generally varied together, and were combined into a single “limb” power.

Results

Mechanical powers for the HAT and the ipsilateral limb, as well as the net mechanical power, are presented in Fig. 1 for the older children. These data represent the powers induced by the hip joint moment (Fig. 1A), knee joint moment (Fig. 1B), and ankle joint moment (Fig. 1C). Note that the net power in Fig. 1 is equivalent to the joint power obtained from an inverse dynamics analysis [3]. Results for the two younger groups were nearly identical to the older group, except that some peak power values were greater in the younger children.

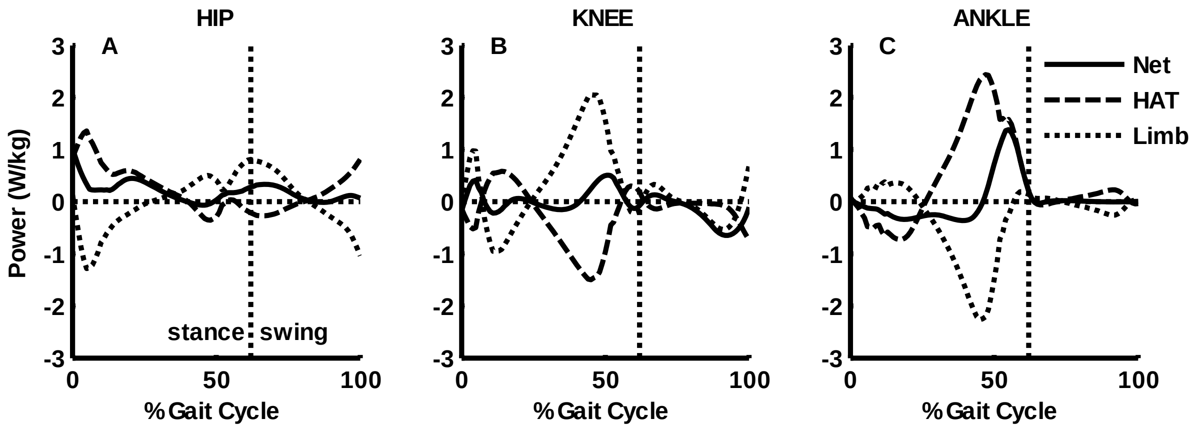


Fig. 1. Net, HAT, and limb powers induced by the hip, knee, and ankle joint moments.

Discussion

During the first half of stance, the hip moment generated energy to the HAT, while also transferring energy from the limb to the HAT (Fig. 1A). The knee moment also transferred energy from the limb to the HAT during the first half of stance, except during initial contact (Fig. 1B). Beginning at about mid-stance, the ankle transferred energy from the limb to the HAT, while also absorbing energy from the limb (Fig. 1C), whereas towards the end of stance, the ankle moment generated energy directly to the HAT (Fig. 1C). During the second half of stance, the knee moment had an energetic effect opposite to the ankle moment (Fig. 1B & C). Overall, these results were in general agreement with earlier data from adults [4]. Among the three age groups, the largest differences were greater peak HAT and limb powers (relative to body mass) due to the hip and knee joint moments in the younger children. For the knee, the greater peak power values were due primarily to greater energy transfer from the HAT to the limb, while for the hip there was both greater energy transfer and more work done on the HAT segment in the younger groups. These differences reflect a greater reliance on more proximal joints in the younger children to produce locomotion.

References

[1] Fregly BJ, Zajac FE. J Biomech 1996; 29:81-90.
[2] Neptune RR, et al. J Biomech 2001; 34:1387-1398.
[3] Robertson DGE, Winter DA. J Biomech 1980; 13:845-54.
[4] Siegel KL, et al. Gait Posture 2004; 19:69-75.
[5] Umberger BR, et al. NACOB, Ann Arbor, MI, 2008.

Acknowledgements

This work was supported by grants 710P and 710U from Kosair Charities, Inc.

First Order Forward Dynamic Gait Simulations

Shawn Russell[1], Bradford Bennett [1,2], Pradip Sheth [1], and Mark Abel[2]

University of Virginia, Charlottesville, VA, USA
[1] Department of Mechanical and Aerospace Engineering
[2] Department of Orthopedic Surgery
E-mail: sdr2n@virginia.edu

INTRODUCTION

Stable walking, in particular the balance and postural stability during gait, has been widely studied in robotics. However, only a few studies have used angular momentum about the center of mass (CoM) bipedal gait[1]. Popovic[2] demonstrated the regulation of angular moment of the full body around its CoM and presented a control strategy where the system angular momentum was explicitly controlled[3]. This strategy was however only presented for the single support phase of walking.

In this study we developed a simple control to simulate walking with a full body patient specific model, limited to the sagittal plane. The model uses segmental angular momentum in a PID (Proportional, Integral, Differential gains) feedback control to develop joint torques to drive the models motion. This control offers inherent global model stability for gait trajectories that it develops, and because the control is non-holonomic it allows the model to be utilized as a predictive device.

CLINICAL SIGNIFICANCE

The development of patient specific forward dynamic models using angular momentum as a control represents a viable clinical evaluation tool. The effects of interventions on pathologic gait can then be predicted by applying characteristic constraints of those interventions to the walking model. Simulations are then able to predict the effect of given interventions on any given specific patient. The global stability of the control also allows for prolonged simulations which allow the model to develop a new steady state gait trajectory.

METHODS

Clinical gait kinematics of 8 test subjects were used to develop/validate the patient specific models. Six of these were without history of lower extremity pathology, one with bilateral paraplegic spastic cerebral palsy, and one with an incomplete C5-C6 spinal cord injury presenting as a slight bilateral crouch gait and drop foot on the left side. All kinematics were collected using Vicon motion analysis system (Oxford, UK). From this data the angular momentum about the CoM for each body segment was calculated and averaged for each subject.

For each subject, a full body model was created in MSC.Adams (MSC.Software) using LifeMod (Biomechanics Research Group, Inc). Models were made patient-specific using subject anthropometric data. The models were composed of 16 segments and 15 joints representing a full body in three dimensions though motion was constrained to the sagittal plane for these simulations.

First order forward dynamic simulations were run using the angular momentum about the CoM of each segment, as a negative feedback control. At each subsequent time point in the gait cycle standard proportional and differential (PID) feedback gain were used to determine individual joint torques applied during the simulation to minimized error between desired and simulated angular momentum.

RESULTS/VALIDATION

Forward dynamic simulations were successfully completed using the subject-specific human model and PID control for both the single and double support phases of gait. Figure 1 shows agreement between the clinically measured joint angles and the forward dynamic solution for a single subject. Results for both normal and pathologic subjects demonstrated similar agreement. Note that small difference in joint angle trajectories in the ankle joints represent the model adapting the gait trajectories to adapt to delayed heel contact.

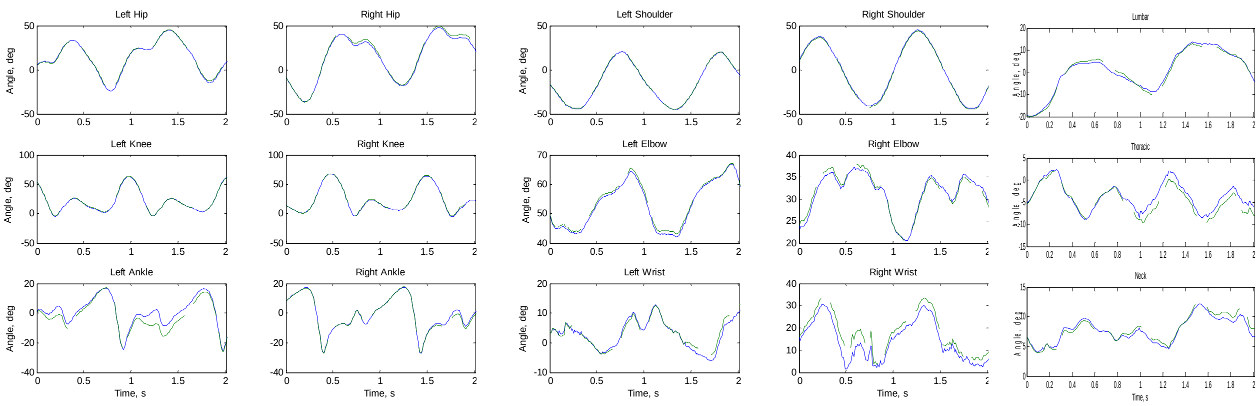


Figure 1. Simulated, dashed, and experimental, solid, joint angle trajectories for a normal gait pattern

CONCLUSIONS

First order forward dynamic simulations demonstrated global stability by controlling the angular momentum of each segment about the total body CoM. The body is in effect stabilized; rotation about the CoM is controlled, and it can not fall. This is the same premise applied in ZMP controls often used for robotics. This contrasts with simulations implementing forward dynamic controls and inverse dynamic feed back controls, which often require external constraints to ensure stability.

This first order global stability comes at a cost, the decreased stability of the individual joint angle trajectories. Because of this the first order forward dynamic simulations are flexible enough to allow the model to adapt its motion from the motion used to develop the control signal. By controlling the angular momentum of the system we can take advantage of the reduced stability/rigidity of joint angle trajectories. For any given control signal the model is able to accommodate small changes, i.e. delayed foot contact, small torque errors etc., develop new joint trajectories and continue walking in a stable gait pattern, thus predicting changes in gait trajectories based on applied constraints.

REFERENCES

1. Goswami, A., Kallen, V. (2004). *Proc IEEE ICRA'04,* 3785-3790
2. Popovic, M. et al. (2004a). *Proc IEEE ICRA'04,* 2405-2411
3. Popovic, M. et al. (2004b). *Proc IEEE/RSJ IROS'04,* 1685-1691

ACKNOWLEDGEMENTS

The authors would like to thank the staff at the Gait and Motion Analysis Lab, Kluge Children's Rehabilitation Center at the University of Virginia, where experiments are conducted. This work was supported in part by NSF grant 0503256.

Gait Analysis for Identification of Lower Extremity Alignment Issues

Jean Stout, PT, MS and Tom Novacheck, MD
Gillette Children's Specialty Healthcare, St. Paul, MN USA

This study demonstrates identification of deformity not previously incorporated into clinical history which significantly altered treatment recommendations and subsequent intervention. He presented to Gait Lab for study in 11/2005 by a local orthopaedist.

Patient History

MN is a 54 year-old male with a diagnosis of a partial avulsion of the semimembranosus from the ischium confirmed by MRI findings. He sustained the injury in 7/2003 while playing racquetball. Since the injury he has had persistent pain and loss of athletic activity. MN and his physician are considering surgical repair want to be confident that the hamstring is the primary issue disrupting his activity. He has a long history of athletic injuries dating back to 1968 when he sustained a left boot-line distal tibial fracture. He notes his foot persistently turned in after the fracture. He has had multiple left ankle sprains. He has a history of a T7 vertebral fracture in 1984 after falling from a scaffold. MN has ad previous arthroscopy surgery of the right knee in 2002 with a lateral meniscotomy.

MN is actively involved in sports and is a personal trainer by profession. His primary frustration is his difficulty in remaining physically active. He is no longer able to run due to pain, unable to roller blade because of marked deterioration (lateral wear pattern) of the forward most wheel on the left after a short distance. On a daily basis he has difficulty ascending and descending stairs (gluteal pain at the ischium) and walking long distances on a hard surface.

Clinical Data

Restricted left ankle dorsiflexion to neutral with both knee flexion and extension. Weakness by manual muscles testing on the left in the following muscles or muscle Groups: hip extensors (grade 4+), adductor (grade 4); hamstrings (grade 4+), and plantarflexors (grade 4+). MN has ischial pain with hip extension. A 1.5 cm leg length difference is noted right longer than left. A -10° thigh foot angle and a 0° bimalleolar axis on the left suggest an internal tibial torsion. A weight-bearing lower extremity x-ray demonstrates bilateral genu varum, and confirms leg-length difference measured on clinical exam. The old left tibial fracture is noted - healed with varus alignment resulting in distal tibial varus (varus ankle mortis). Pes cavus foot deformity is noted bilaterally with increased stiffness.

Gait Data:

Video review demonstrates an atypical degree of supination of the left foot.

Kinematic data using a functional model confirm the internal tibial torsion measured by clinical examination exhibiting an internal transcondylar vs. bimalleolar axis on the left. Left foot progression angle is internal (Figure 1). The remainder of the kinematic data is within typical limits.

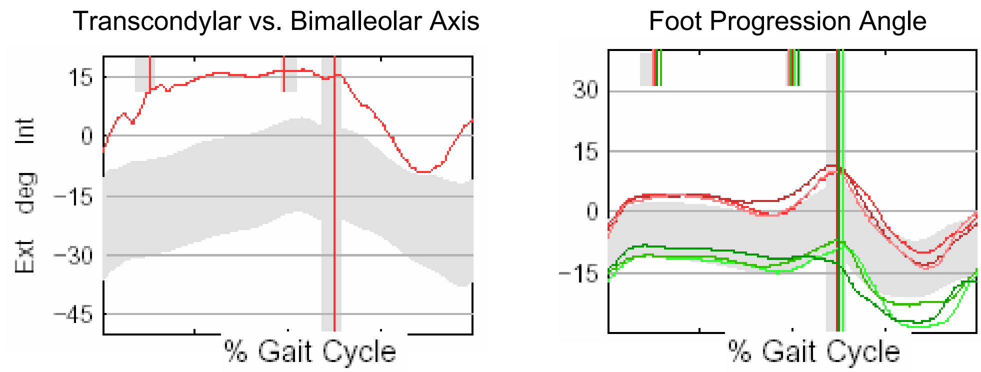
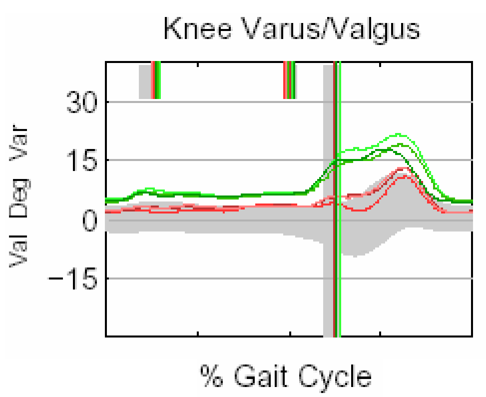


Figure 1: Kinematic Data

Kinetic data: Sagittal plane kinetic data are within typical limits. Coronal plane kinetic data demonstrate an increased valgus moment consistent with lateral knee joint stress bilaterally (Figure 2).



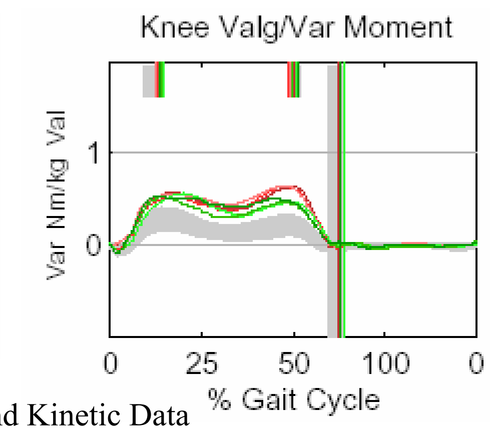


Figure 2: Coronal Plane Kinematic and Kinetic Data

Treatment Decisions and Indications:

The combination of the leg length discrepancy, internal malrotation (internal tibial torsion) and high-arched foot alters the shock absorbing function of the left lower extremity. The distal tibial varus along with the mal-alignment issues likely contribute to persisting ankle sprains and the prolonged symptoms relating to the semimembranosus tear. Surgical repair of the semimembranosus tear may not lead to relief of symptoms. Future function of the knee and ankle may be at risk. Treatment options could include use of a shoe lift and shock absorbing orthotics or lower extremity realignment surgery to preserve long term knee and ankle joint function.

Summary:

Gait analysis is useful in identifying alignment and mechanical issues that contribute to underlying pathology and daily function that may be not clearly identified by clinical exam and history alone. Objective quantitative evidence to augment clinical data is obtained and an alternate treatment plan is proposed.

INFLUENCE OF GAIT ANALYSIS ON DECISION MAKING FOR HIP ADDUCTOR LENGTHENING SURGERY

Tishya A. L. Wren, Anthony A. Scaduto, Richard E. Bowen, Norman Y. Otsuka, Reiko Hara, Minya Sheng, Linda S. Chan, Robert M. Kay
Childrens Hospital Los Angeles & Shriners Hospital for Children, Los Angeles, CA

INTRODUCTION

Several studies have shown that clinical decision making is altered when gait analysis data are considered[1-4]. However, a weakness of these studies is their reliance on observational cohorts, which are not controlled for potential confounders such as surgeon uncertainty. Reported results differ widely among studies, with gait analysis changing recommendations for hip adductor lengthening surgery in 3-60% of cases[1-4]. The current study used data from a randomized, controlled trial (RCT) to examine in greater detail the influence of gait analysis data on decision making for hip adductor lengthening surgery (ADD).

Our main hypothesis was that decision making for ADD would be influenced by recommendations received in a gait analysis report. Specifically, we hypothesized that: (1) Surgeons are more likely to proceed with planned ADD surgery if they receive a gait report supporting this plan; (2) Surgeons are more likely not to proceed with planned ADD surgery if they receive a gait report recommending against this plan; (3) If ADD is not planned initially, it will be performed more often when the surgeon receives a gait report recommending ADD; (4) If ADD is not planned and the gait report agrees that it should not be done, receiving a gait report will have no effect because ADD will rarely be done.

CLINICAL SIGNIFICANCE

These results from a randomized, controlled trial provide a stronger level of evidence demonstrating the impact of clinical gait analysis on surgical decision making. This impact includes reinforcement of planned ADD surgery when a gait report supporting this treatment is available.

METHODS

This study included 130 ambulatory children with cerebral palsy (CP), age 3-18 yr, who were candidates for lower extremity orthopaedic surgery to improve gait. All subjects underwent pre-operative gait analysis including physical examination, computerized gait analysis, and electromyography, and standard clinical gait reports were produced. The subjects were randomized to two groups. For subjects in the Treatment group, the referring surgeon received the patient’s gait analysis report; for subjects in the Control group, the referring surgeon did not receive the gait report.

Data on ADD and other surgeries was collected at three time points: 1) referral by treating surgeon before gait analysis, 2) recommendations by gait laboratory surgeon after gait analysis, 3) actual surgery done. The unit of analysis was patient-side. For unilaterally involved subjects, only the affected side was included. The main outcome measure was the relative acceptance between the Treatment and Control groups, where acceptance is defined as the likelihood that the referring surgeon accepted the gait analysis recommendation for ADD. Statistical significance was determined using the 2-sided Fisher's exact test.

RESULTS

Of those sides that had ADD planned by the referring surgeon before gait analysis, the vast majority also had ADD recommended by the gait laboratory (*Both Referral & Gait Lab*) (Table 1). ADD was ultimately performed in all of the sides (24/24) in subjects receiving the gait report (*Treatment*), but only 79% (15/19) of the sides in subjects not receiving the report (*Control*), relative acceptance: 1.27 (95% CI: 1.00, 1.60; p=0.031). These results suggest that the gait report reinforced the referring surgeons' original decision making, increasing their confidence to proceed with the procedure.

The gait report also influenced decision making in the few cases where the gait laboratory disagreed with the referring surgeon's plan to do ADD (*Referral Only*). The treatment plan was changed for all sides in patients receiving the gait report (3/3), but not for the only side in a patient without the gait report (0/1) (relative acceptance undefined; p=0.25).

The gait laboratory recommended ADD in a large number of sides that did not have ADD planned before gait analysis (*Gait Lab Only*). ADD was performed in 15% (4/26) of sides in subjects who received the gait report, compared with 13% (4/32) of those not receiving the report, relative acceptance: 1.23 (95% CI: 0.34, 4.45; p=1.00).

For those sides where the surgeons did not plan and the gait analysis did not recommend ADD (*None*), this plan was followed in 53/55 (96%) of sides in the Treatment group and 56/58 (97%) of sides in the Control group, relative acceptance: 1.00 (0.93, 1.07; p=1.00).

Table 1: ADD surgery at each time point for Treatment and Control groups

<table>
<tr><td colspan="4">Planned Before GA
27/108 (25%)
20/110 (18%)</td><td colspan="4">Not Planned Before GA
81/108 (75%)
90/110 (82%)</td></tr>
<tr><td colspan="8">***Treatment Group (N=108)***
Control Group (N=110)</td></tr>
<tr><td colspan="2">*Both Referral & Gait Lab*</td><td colspan="2">*Referral Only*</td><td colspan="2">*Gait Lab Only*</td><td colspan="2">*None*</td></tr>
<tr><td colspan="2">Recommended by GA
24/27 (89%)
19/20 (95%)</td><td colspan="2">Not Recommended by GA
3/27 (11%)
1/20 (5%)</td><td colspan="2">Recommended by GA
26/81 (33%)
32/90 (36%)</td><td colspan="2">Not Recommended by GA
55/81 (67%)
58/90 (64%)</td></tr>
<tr><td>Done
24/24 (100%)
15/19 (79%)</td><td>Not Done
0/24 (0%)
4/19 (21%)</td><td>Done
0/3 (0%)
1/1 (100%)</td><td>Not Done
3/3 (100%)
0/1 (0%)</td><td>Done
4/26 (15%)
4/32 (13%)</td><td>Not Done
22/26 (85%)
28/32 (87%)</td><td>Done
2/55 (4%)
2/58 (3%)</td><td>Not Done
53/55 (96%)
56/58 (97%)</td></tr>
</table>

DISCUSSION

The randomized design allowed us to demonstrate a reinforcing effect of gait analysis that has not been previously reported. In a companion study using the same methodology, we found similar results for hamstring lengthening surgery. This controlled design has the potential to provide stronger evidence on the clinical impact of gait analysis. Data collection is ongoing.

REFERENCES: [1] DeLuca et al., *J Ped Orthop* 17:601-14, 1997. [2] Kay et al., *Clin Orthop* 372:217-22, 2000. [3] Cook et al., *J Ped Orthop* 23:292-5, 2003. [4] Lofterod et al., *Acta Orthop* 78:74-80, 2007. Support provided by AHRQ grant # 5 R01 HS014169.

Metabolic Analysis of Gait for Decision-Making Regarding Mode of Community Mobility

Nancy Lennon MS, PT, Chris Church MPT, Daveda Taylor DPT,
Stephanie Marchesi PhD, John Henley PhD, Freeman Miller MD
Alfred I. duPont Hospital for Children Wilmington, DE, USA

PATIENT HISTORY: This is a case review of an 18 year-old young man, with spastic diplegic pattern CP; GMFCS level III. The objective of this case study is to examine the changes in the metabolic efficiency of his gait from age 11 to age 18. This boy used a walker for his primary mobility from age 2 to 10 and transitioned to forearm crutch use at age 10. He reported his primary community mobility as crutch walking up to age 14. Previous orthopedic surgery included, at age 3, bilateral adductor lengthening, bilateral iliopsoas lengthening, bilateral distal hamstring lengthening and bilateral rectus femoris transfers and, at age 6, right tibial derotation osteotomy, bilateral gastrocnemius recessions, and bilateral distal hamstring lengthening. Previous spasticity management included baclofen pump implantation at age 11 with gradually increased dosages over a three year span.

CLINICAL DATA: Table 1 provides information from visits at age 11 (to assist with baclofen pump decision) and at age 14 (to determine further baclofen treatment or discontinuation). During the first 2 years of baclofen pump use, this young man reported improved ability to negotiate stairs and improved independence in ADLs, especially dressing. At age 14, during his first year of high school, he reported increasing difficulty getting around with crutches, and more fatigue.

Table 1 Data from clinical exams at age 11 and age 14

R / L	**BMI**	**hip ext**	**knee ext**	**pop angle**	**TMA**	**MMT**	**LE Ashworth**	**GMFMD (%)**	**LE Control**
Age 11	22.5 (92%)	-23 -27	-9 -7	55 52	21 ext 29 ext	3- to 3+	2 – 3's t/o	69%	Poor to Fair
Age 14	27.7 (>95%)	0 0	+10 + 8	60 60	30 ext 27 ext	2 to 2+	0 - 1's t/o	28%	Poor to Fair

GAIT DATA

Temporal-Spatial Data: From age 11 to age 14 there was unchanged walking speed (78 to 80 cm/sec) decreased walking cadence (123 to 93 steps/min) and increased stride length (75 to 100 cm).

Kinematic Data: Deviations were numerous and similar at both visits. While walking with forearm crutches there was exaggerated lateral trunk lurch at age 14, and excess pelvic rotation, forward trunk lean and anterior pelvic tilt at both visits. There was hyper hip flexion during swing phase and knees had a rapid extensor thrust during stance phase at both visits. There was external tibial torsion bilaterally and dynamic equinous at age 11 with reduced first and second ankle rockers at age 14.

Metabolic Data: Table 2 provides data from metabolic testing. O_2 cost index is a z-score comparing the child to a size predicted value and VO_2 reserve calculates the difference between peak and walking VO_2 for an indicator of aerobic capacity.

Table 2 Metabolic Data for walking with forearm crutches at age 11 and 14

	HR Rest	HR walk	VO_2 rest	VO_2 walk	VE Walk	O_2 Cost	O_2 Cost Index	Peak VO_2 Exercise	VO_2 Reserve
Age 11	67 bpm	125 bpm	4.7 ml/kg/min	17.0 ml/kg/min	30 l/min	0.62 ml/kg/m	+5.3	*Not Tested*	*Not Tested*
Age 14	90 bpm	140 bpm	3.6 ml/kg/min	20.3 ml/kg/min	50.4 l/min	0.58 ml/kg/m	+5.8	21.0 ml/kg/min	0 %

TREATMENT DECISIONS AND INDICATIONS

Following the gait analysis at age 14 there was a discussion among the patient, family, orthopedic physician, and physiatrist. The patient's goals were to "hang out" with his friends and learn to drive a car. Parents' goals were for his long-term independence and weight management. The following treatment plan was established.

Due to complaints of fatigue, reduction in muscle strength, and deterioration in gross motor function (by GMFM), a recommendation was made to wean from the baclofen for a trial period. Due to the very high energy cost of walking and the lack of aerobic reserve between walking and exercise, a recommendation was made for manual wheelchair use in the community. Due to the high BMI and concerns of further weight gain and fitness deterioration, a recommendation for participation in a structured exercise program or adaptive sports was made.

At age 18, this young man was invited back for metabolic testing as part of this case study. The baclofen pump was removed at age 14. This young man now uses a manual wheelchair for his primary mobility. He walks around the house and when necessary in the community to overcome limited w/c accessibility. He is very active in semi-professional sports including wheelchair basketball and wheelchair rugby. He reports some problems with LE spasticity during sports. He drives a car with hand controls and manages his manual wheelchair independently (except for curbs). His BMI is 29.1 (95%)

Table 3 Metabolic Data for walking with forearm crutches at age 18

	HR Rest	HR walk	VO_2 rest	VO_2 walk	VE Walk	O_2 Cost	O_2 Cost Index	Peak VO_2 Exercise	VO_2 Reserve
Age 18	83 bpm	120 bpm	3.0 ml/kg/min	11.6 ml/kg/min	59 l/min	0.33 ml/kg/m	+2.6	16.3 ml/kg/min	29%

SUMMARY

Metabolic analysis demonstrated a very high cost of walking at age 14 that was not conducive to full community ambulation. The lack of aerobic reserve as measured by exercise testing further demonstrated the impracticality of community ambulation for him. This young man's choice of wheelchair use as a more energy efficient mode of community mobility allowed him to use his aerobic reserve for sports participation, at which he now excels. His case demonstrates that fitness deterioration and weight gain are not requisite consequences following transition to a wheelchair for an adolescent with CP. This young man's participation in sports allowed him to maintain and even improve his level of aerobic fitness. At 4-year follow-up he has improved his aerobic reserve (VO_2 reserve) and has lowered his energy cost of walking.

Intralimb Coordination during Gait: Inter Model Comparison

Arvind Ramanujam [1], Gail F Forrest [1], Sue Ann Sisto [1,2]
[1]Kessler Medical Rehabilitation Research and Education Center, West Orange, NJ, USA;
[2]Department of Rehabilitation Science, Stony Brook University, Stony Brook, NY
E-mail: aramanujam@kmrrec.org Web: www.kmrrec.org

Introduction

Intralimb and interlimb coordination techniques have previously been used to review gait in able body as well as pathological gait.[1-3] The objective of this study was to examine the reliability, validity, and variability of four Plug-In-Gait models that differ computationally (VICON, Oxford Metrics, Oxford, UK) using the three-dimensional intralimb coordination profiles for the hip, knee and ankle segments. Continuous relative phase (CRP) time series analyses compared intra limb coordination patterns within each model.[3] Root-mean-square (RMS) analyses determined magnitude changes for the CRP.[3]

Clinical Significance

The Plug-in-Gait model is a widely accepted biomechanical model for evaluating gait dynamics and has been extensively used as a human diagnostic tool by clinicians and physicians internationally. It is therefore important to be well versed with the various intricacies of the model before making clinical interpretations.

Methods

Able bodied gait data was collected at 120Hz using a Vicon motion capture system from eight participants who provided informed consent. Markers were placed on anatomical landmarks as defined by Plug-in-Gait. Anthropometric measurements were recorded manually. Data collection included four different static protocols (Table 1) and ten dynamic gait trails. Five trials (left leg only) were chosen for data analysis. Data was filtered (6 Hz low pass butterworth filter) and normalized to percent gait cycle (Matlab, The MathWorks, Natick, MA). CRP and RMS analyses were computed using angular position (θ) - velocity (ω) phase planes for each segment normalized to a unit circle.[3] To maintain inter and intra subject reliability all data collection and analyses using the Plug-in-Gait model were performed by the same tester.

Table 1. Description of the four static models (SM) compared

SM 1	SM 2	SM 3	SM 4
Plug-in-gait marker set without KAD	Use of Knee Alignment Device (KAD)	Using medial knee/ankle markers and shank rotation body builder model	Using the medial ankle markers and the KAD

Results

The four models determined bi-planar differences in the coordination profiles for the hip, knee and ankle (Figure 1-b,c). The sagittal plane profiles were unaffected (Figure 1a). Mean root-mean-square differences (MRMSD) and standard deviations (SD) for intralimb couplings (Hip.Knee, Knee.Ankle, Hip.Ankle) were calculated for all models in all three planes (Figure 1-d,e,f). Sagittal plane MRMSD (85.87±0.12˚) and SD (±7.25˚) remained unchanged (Figure 1d). Greater variability in MRMSD (83.49±6.99˚) with consistent SD (±19.56˚: SM1 to SM4) occurred in the transverse plane (Figure 1f) while greater variability in SD for SM3 (±33.94˚) occurred in the frontal plane (Figure 1e) with minimal SD's for SM2 (±6.91˚) and SM4 (±10.09˚). SM2 and SM4 performed consistently with similar MRMSD in terms of magnitudes with lower standard deviations in all three planes.

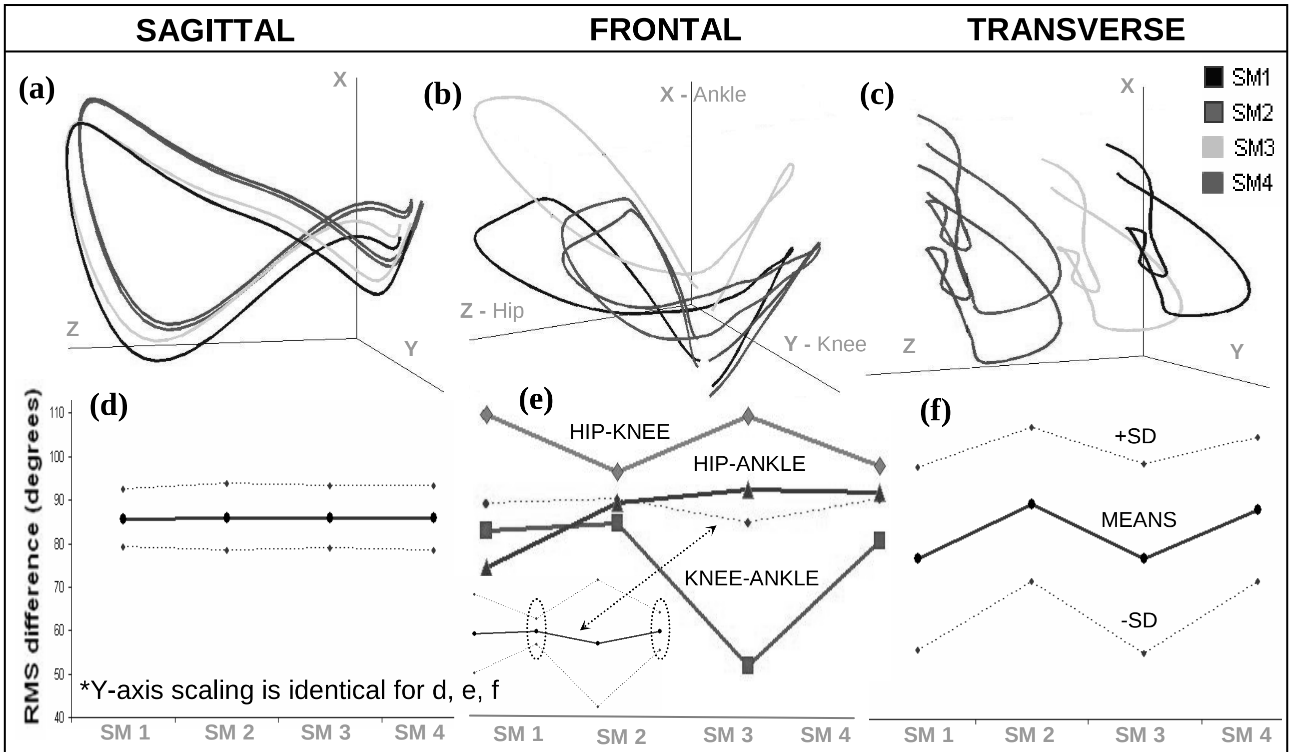


Figure 1. (a)(b)(c): 3D kinematic Hip.Knee.Ankle coordination profiles for the four models; (d)(e)(f): Mean root-mean-square differences (MRMSD) between intralimb couplings (individual profiles shown for frontal plane 1e)

Discussion

Different computational models generate different coordination profiles, especially in the frontal and transverse planes. These deviations in coordination are directly related to the different algorithms implemented to calculate knee and ankle joint centers, leading to different segmental coordinate systems. In summary, our results demonstrate that kinematics and segmental coordination profiles are directly linked to the model. The variability in the MRMSD and SD across all models (especially SM3 in the frontal plane) ultimately questions the reliability and validity of the clinical report. Clinicians need to understand the bias and limitation attached to these models. More work needs to be done in these inter model comparisons to establish validity.

References

1. Whitall, J., et al (1992) Journal of Motor Behavior 24:339-53
2. Byrne, J.E., et al (2002) Perceptual and Motor Skills 94:265-80
3. Haddad, J.M., et al (2006) Gait and Posture 23:429-434

INDIVIDUALS WITH PARKINSON'S DISEASE EXHIBIT A DECREASE IN DYNAMIC STABILITY DURING WALKING COMPARED TO THE ELDERLY AND YOUNG

Christopher J. Arellano[1] and Max J. Kurz[2]

[1]University of Colorado, Department of Integrative Physiology, Boulder, CO
[2]Munroe-Meyer Institute for Genetics and Rehabilitation, University of Nebraska Medical Center, Omaha, NE

INTRODUCTION

One of the hallmark features of Parkinson's disease (PD) is its affect on walking. Individuals with PD often walk at a slower pace, have a shorter stride length and have an increased amount of stride-to-stride variability [1]. Many of these changes during walking have been speculated to be attempts to prevent falls. However, 70% of patients with PD experience an annual fall [2]. Clinical metrics that adequately measure fall susceptibility during walking remain elusive despite the considerable amount of experimental work on walking and postural balance of individuals with PD. Floquet analysis has recently been used to assess the walking stability impairments observed in the elderly [3]. In this analysis, dynamic stability of the walking pattern is based on the rate at which the state variables of the system (*i.e.*, angular displacements and velocities) return back to a steady-state periodic walking pattern after experiencing a disturbance that can be due to an external perturbation or a neuromechanical error. The walking pattern is considered to have greater dynamic stability if it returns back to the steady-state walking pattern at a faster rate. The purpose of this investigation was to determine if Floquet analysis discriminates between the dynamic stability of individuals with PD, who have a history of falls, and the elderly and young, who have no fall history. We hypothesized that individuals with PD, who have a greater number of annual falls, will exhibit a decrease in dynamic stability while walking compared to a group of healthy elderly and young individuals.

CLINICAL SIGNIFICANCE

The results presented here indicate that Floquet analysis may provide a means for differentiating the effects of age and PD on the dynamic stability of the walking pattern. These results have implications for identifying individuals who have greater fall susceptibility. Since majority of falls occur during walking, Floquet analysis may supplement current clinical measures that assess balance in patients with movement disabilities.

METHODS

Healthy young adults (N=5; Age = 22.6 + 2.2), elderly (N=5, Age = 74.2+5.8), and individuals with PD (N=5; Age= 74.0 + 4.8) participated in this study. All the aged individuals were healthy, active and had no history of recent falls. The individuals with PD had a Hoehn and Yahr scale score that ranged between 2-3, and had a history of recent annual falls (range 1-36). A six-camera motion capture system (200 Hz) was used to collect the right leg's ankle, knee and hip joint sagittal plane kinematics as the subjects walked on the treadmill at a self-selected pace for 3 minutes. The PD participants performed the walking trial while on their levodopa medication. All participants walked while holding onto the handrails of the treadmill.

Floquet analysis was used to quantify the dynamic stability of the walking pattern [3]. A state vector (**x**) was defined by the joint angular position and velocity of the hip, knee, and ankle at heel-contact. Disturbances in the walking dynamics were represented as a nonlinear map of the state vector at stride *n* to *n+1* and expressed as

$$\delta \chi^{n+1} = \mathbf{J}\,\delta \chi^{n} \qquad \text{Eq. 1}$$

where δ denotes the deviation about **x** at each n[th] stride. **J** is the Jacobian which represented the evolution of the state vector from one stride to the next. δ**x** at each heel-contact were calculated based on the difference between **x** at stride *n* and the mean value of **x** across 100 strides. A least squares algorithm was used to solve for **J**. The largest eigenvalue (β) was computed from **J** and used to quantify the stability of the walking gait pattern. A β value further away from zero signified a less stable walking gait pattern [3].

RESULTS

All three groups were similar with respect to height ($p = 0.58$) and weight ($p = 0.15$). The PD and elderly group were similar in age ($p > 0.05$). The young (0.88 ± 0.24 m/s) and elderly (0.95 ± 0.15 m/s) walked significantly faster ($p < 0.05$) than the PD group (0.52 ± 0.09 m/s). The young and elderly walked at similar speeds ($p > 0.05$). As seen in Figure 1, β was significantly greater for the PD group compared to the young ($p < 0.05$) and elderly ($p < 0.05$). There were no significant differences in β between the young and elderly ($p > 0.05$).

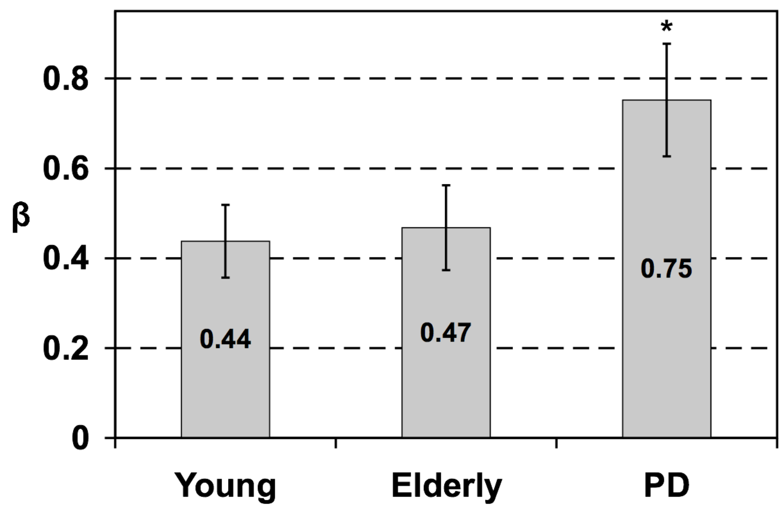


Figure 1. β values for each group (mean ± SD). * indicates that β was significantly greater for the PD group (p < 0.05) compared to the young and elderly.

DISCUSSION

Our results signify that individuals with PD have a less stable walking pattern. The larger β seen in the participants with PD indicates that it takes them longer to correct for disturbances that occur during walking. The inability to quickly respond to disturbances may be the reason why individuals with PD had a higher incidence of falls. The β values for the PD participants in this study were similar to what was previously reported for a group of elderly with a history of falls [3]. We suggest that Floquet analysis may provide a means for assessing fall susceptibility. However, a study of a larger patient population with various levels of balance disorders is necessary to fully support this initial claim.

REFERENCES

[1] Morris et al. (2001). *Clin Biomech* 16:459-470; [2] Bloem et al. (2003). *Curr Opin Neurol* 16:15-26; [3] Granata & Lockhart (2008).18(2):172-8.

GAIT IN CHILDREN WITH AUTISM AND ASPERGER'S DISORDER

McGinley, J., PhD[1,] Murphy, A., PhD[2,3], Rinehart, N., PhD[3], Tonge, B., PhD[3], Bradshaw, J., PhD[3], Iansek, R., PhD[1]

[1] Gait CCRE, Murdoch Childrens Research Institute, Melbourne, Australia, [2] Kingston Centre, Southern Health, Melbourne, Australia, [3] Monash University, Melbourne, Australia

INTRODUCTION

Autism and Asperger's disorder are prevalent, debilitating neurodevelopmental disorders, characterised by impaired communication, social difficulties and the presence of repetitive behaviours or circumscribed interests. The number of individuals diagnosed with autism and AD has increased markedly over the last decade. Clinical descriptions of people with both disorders often include mild or subtle gait or postural abnormalities. The empirical gait data available is limited, and the diagnostic criterion for subject inclusion in some studies is unclear. Previous studies have suggested that gait is more variable in people with autism [1,2,3]. Clinical observations have reported specific kinematic abnormalities such as toe walking [4], reduced ankle range of motion [1] and abnormalities of trunk or arm posture [3]. The aim of this study is to quantify and characterise the full-body kinematic features of gait associated with high functioning autism (HFA) and AD.

CLINICAL SIGNIFICANCE

Disordered movement is a clinical feature of both HFA and AD. Understanding whether the motor disorders associated with autism and AD are similar or different will help define if autism and AD are on a symptom continuum, or are distinct disorders. Quantitative gait analysis can provide objective evidence of the nature of any gait disorders to further inform the clinical descriptions.

METHODS

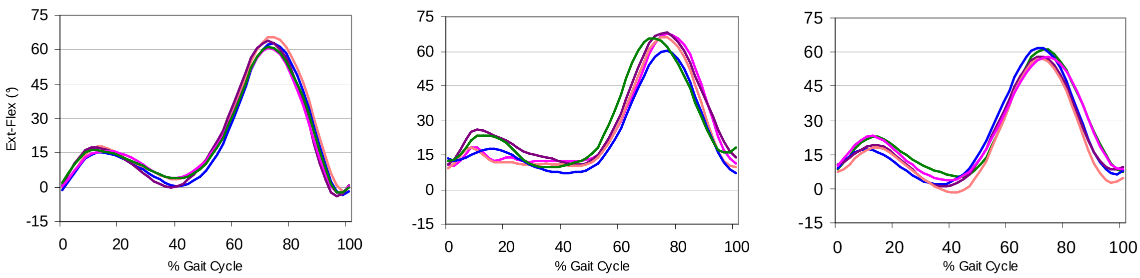
Three-dimensional full body gait data were captured from a reference group of normally intelligent children (n=9, 11.0±0.6years), a group with high functioning autism (HFA) (n=9, 10.5±1.2years) and a group with AD (n=9, 10.6±1.1years). All participants completed five walking trials at self-selected preferred, fast and slow speeds. A conventional biomechanical model (Plug-in-gait) was used to calculate full body joint kinematics for 2 left and 2 right strides of each trial. This preliminary analysis selected kinematic variables of interest including peak values and range of motion (ROM) of the shoulders, trunk and lower limb joints and the centre of mass (COM). Within-subject inter-trial variability of the kinematic curves in each condition was examined. One –way ANOVA was used to investigate between-group conditions in each gait condition.

RESULTS

At preferred speed, no significant between-group differences in the selected kinematic variables were evident. During fast walking, significant ($p < .05$) group differences emerged for peak trunk tilt, trunk tilt ROM, and peak shoulder abduction. Vertical

ROM of the COM also approached significance (p = .054). Post-hoc analyses showed that, relative to the reference control group, the children with HFA walked with more trunk peak flexion, increased range of trunk flexion/extension, greater peak shoulder abduction and greater vertical ROM of the COM. Differences between the children with AD and the reference group also approached significance for peak trunk flexion and peak shoulder abduction.

Inspection of inter-stride variability found that the HFA group were most variable for almost all kinematic variables across the three speeds. At preferred speed, significant differences were found, relative to the control group, in variables including shoulder abduction, pelvic tilt, knee flexion, and ankle dorsiflexion. At slow speeds, both the HFA and AD group were more variable at the knee and ankle than the reference group. Children with HFA were most variable in COM vertical oscillations in both the slow and preferred speed walking conditions. Figure 1 shows examples of the inter-stride variability of knee flex/extension data from the right side of (left): a child from the reference group, (centre): a child with HFA, and (right): a child with AD.



DISCUSSION
These preliminary findings are consistent with some of the clinical observations of HFA and AD gait and support previous studies suggesting that upper body and upper limb abnormalities may be key features of the subtle gait impairments in both disorders. They are also broadly consistent with previous studies suggesting that children with HFA have a more variable gait pattern than those with AD. The results confirm that alternate speed conditions such as fast or slow walking may be useful as a more sensitive indicator of subtle gait dysfunction. The small sample size and limited statistical power currently precludes definitive conclusions. Further evidence from a larger study sample is needed (and in progress) to further elucidate group differences.

REFERENCES
Hallet et al. 1993 *Arch Neurol* 50; 13041308
Vernazzo-Martin et al. 2005 *J Aut Dev Disorders* 35(1); 91-101
Rinehart et al. 2006 *Eur Child Adolesc Psychiatr* 15(5); 256-64
Weber 1978 *Acta Paedospychiatr* 43; 83-83

A CASE STUDY FOCUSING ON SPEED ADAPTATIONS RECOGNIZED BY GAIT ANALYSIS IN CEREBRAL PALSY

Kaat Desloovere[1,2], **Deborah Severijns**[1,2], **Guy Molenaers**[3]
[1]Clinical Motion Analysis Laboratory of the University Hospital Pellenberg, [2]Department of Rehabilitation Sciences, [3]Department of Musculoskeletal Sciences (Pediatric Orthopedics) , K.U.Leuven, Belgium

Patient history

The patient is a 6+7 year old girl, diagnosed with spastic diplegia (periventricular leucomalacia), GMFCS II. The early milestones were characterized by delayed sitting and crawling and independent walking was achieved at 29 months. She previously received three sessions of multilevel botuline toxine A (BTX-A) treatment, combined with casting, orthotic management and intensive physical therapy (PT) (at 3+2, 4+10 and 5+10 years). She did not have previous lower extremity surgery. She wore posterior leafsprings bilaterally (rigid leafspring/shoe tuned in neutral position) for 100% the day, and KAFOs with an abduction/exorotation rod for 50% of the night . She had 3 to 4 sessions of 30 minutes PT (NDT oriented). Parents were concerned about lost of functionality and lack of progression in therapy. To evaluate the adaptations to speed the child was asked to walk at self selected comfortable walking speed, then at self selected faster speed and finally as fast as possible.

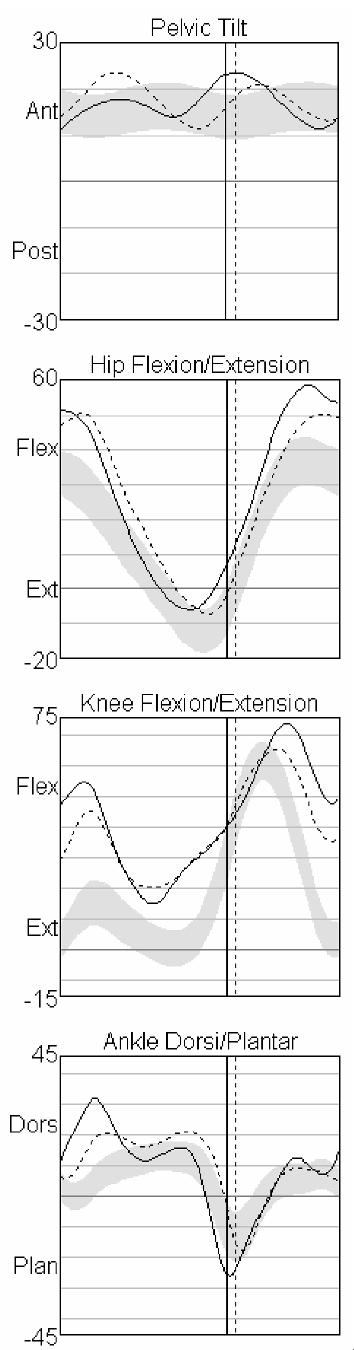


Figure 1: Sagittal pelvic, hip, knee and ankle kinematics for a representative trial of the right (solid) and left (dotted) side. (Normal reference = gray band)

Relevant clinical data

The patient presented with a femoral anteversion of 40° bilaterally, a bilateral hip extension deficit of 10° and popliteal angle of -60° (shift of 5°). She had quite normal ankle ROM bilaterally. Spasticity (Modified Ashworth) was noticed bilaterally for hip flexors (1+ R, 1 L) and adductors (2, R&L), hamstrings (2 R&L), gastrocnemius (3, R&L) and tibialis posterior (1, R&L). She had bilateral ankle clonus and a mild positive Duncan Ely (1, R&L). Decreased strength and selectivity was recognized at the level of the hip, knee and ankle (weakness R>L, manual testing 2- to 3+).

Gait

Visual observation indicated forefoot contact followed by premature heel rise bilaterally, trendelenburg (R>L), and forward trunk leaning. Study of repeated gait trials indicated a lack of consistency in kinematics, kinetics and EMG. Her sagittal plane kinematics were characterized by pelvic anterior tilt with severe double bump, combined with decreased hip extension at terminal stance and knee flexion at initial contact, bilaterally. Both knees showed decreased extension in stance and delayed knee flexion in swing. Ankle kinematics were characterized by a double bump pattern and a reversed second rocker (R>L). (Figure 1) The transverse and coronal plane kinematics showed increased pelvic ROM (with pelvic drop), and internal hip rotation (R>L), as well as a functional leg length difference. She showed a premature ankle plantar flexion moment in early stance and decreased ankle power generation at push off, combined with increased hip power generation (H3), on both sides. Hip abduction moment was reduced, hip extension moment prolonged and knee extension moment increased (R > L). (Figure 2) EMG data at self selected speed indicated rectus femoris activity in swing, cocontraction pattern of hamstrings and rectus, premature activity of gastrocnemius, and lack of tibialis anterior activity at terminal swing. Muscle length graphs at increasing walking speed indicated for an increased lengthening velocity of gastrocnemius at loading response, resulting in a pronounced and earlier reflex activity in EMG (Figure 3). This caused a worsening of the previously described gait deviations at the ankle. Knee flexion at initial contact clearly increased at higher walking speed. Howvere, there was no increased lengthening velocity of hamstrings at terminal swing. EMG data of the proximal muscle groups at higher speeds indicated for a increased cocontraction pattern in hamstrings and rectus femoris.

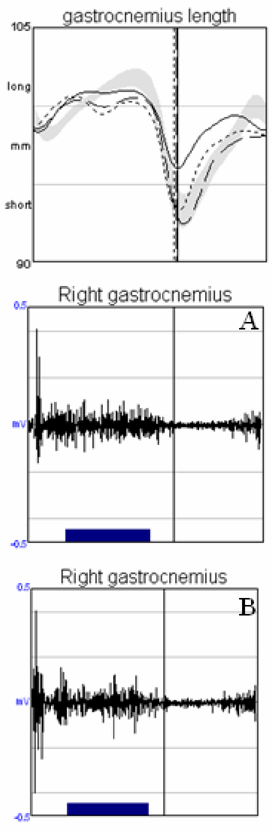


Figure 3: Gastrocnemius length at 3 velocities: self selected (solid), faster (dashed) and fastest (dotted) and raw EMG at self selected (A) and fasted (B) speed.

Figure 2: Hip, knee and ankle moment for a representative trial of the right (solid) and left (dotted) side. (Normal reference = gray band)

Treatment decision

From the gait analysis and clinical examination it was concluded that the patient had pathological gait defined by primary (spasticity) as well as secondary problems (muscle contractures and lever arm dysfunction). Integration of kinematics, kinetics and EMG at self selected speed showed the pathological use of the hamstrings to extend the hip (creating an increased risk of crouch). Furthermore, walking at high speed highlighted increased impact of spasticity in the gastrocnemius and an increased fixation pattern (cocontraction pattern in proximal muscle groups). Although gait deviations were partly caused by secondary problems, the gait pattern was found to be immature, indicating for conservative (reversible) treatment approach. Due to the complexity of the gait pattern, tone reduction treatment, combined with individualized intensive goal oriented PT, should be targeted on the major problem. Within the multidisciplinary discussion it was decided that the pathological use of hamstrings and cocontraction pattern in the proximal muscle groups created the highest risks (crouch) for the future. Integrated treatment, including targeted BTX-A injections, casting, orthotic management and intensive PT was planned to tackle these major problems.

THE INFLUENCE OF WALKING SPEED ON GAIT CHARACTERISTICS IN CHILDREN WITH SPASTIC CEREBRAL PALSY

Severijns D. PT[1], Bruyninckx H. Ir PhD[2], Wambacq H. Ing[2], Aertbeliën E. Ir[2], Fagard K. PT[1], Dan B. MD, PhD[3], Janssens L. Ing[4], Molenaers G. MD, PhD[5] ,Van Campenhout A. MD [5], Desloovere K. PhD[1, 6]

[1] Clinical Motion Analysis Laboratory, University Hospital, Pellenberg, Belgium;[2] Department of Mechanical Engineering, K.U.Leuven, Belgium;[3] Department of Neurology, University Children's Hospital Queen Fabiola, Brussels, Belgium;[4] Groep T, Leuven Engineering School, K.U.Leuven, Belgium;[5] Department of musculoskeletal sciences, K.U.Leuven, Belgium;[6] Department of Rehabilitation Sciences, K.U.Leuven, Belgium

Introduction: Gait parameters in typically developing (TD) children are found to be related to speed.[1, 2] In children with Cerebral Palsy (CP), increased speed may not only augment influence of spasticity, since spasticity is defined as a velocity dependent increase of muscle activity [3], but also increase pathological patterns related to a higher demand of motor control. *Objectives* of the study were to compare kinematics and kinetics, peak muscle lengths and lengthening velocity in CP children with TD children walking at comparable gait speed and to explore differences between TD and CP children in adaptation strategies to higher gait speeds.
Clinical significance: Clinically, gait data are compared with age referenced data of normal population, without taking into account influence of gait speed. Examination of the adaptation strategies used to adapt to higher gait speed, may provide more insight into the influence of spasticity and/or motor control on gait in CP children.
Methods: 14 ambulant children with spastic CP (age 10,8±2,4y, GMFCS I-II, 7 hemi- and 7 diplegia) as well as 4 age-matched TD children were evaluated. The CP group was characterized by mild to moderate spasticity (Modified Ashworth of gastrocnemius and hamstrings=1-3, hip flexors=1-2). Children were asked to walk at self selected comfortable walking speed (CWS), then at self selected faster speed (S1) and finally as fast as possible (S2) without running on a 10-m long walkway. All children received full gait analysis including lower limb kinematics and kinetics (8 camera Vicon system, PlugInGait model, 2 AMTI force plates). Three trials of one random side for each speed condition were used for further analysis. Velocity of muscle lengthening was calculated using custom made software (Matlab®, The Mathworks). Non-dimensional speed according to Hof [4] indicated that CWS of TD children approached S2 of children with CP. Therefore, gait data of the CP group at S2 were first compared to TD children at CWS. Subsequently data of CWS were compared with S2 in both groups to find out whether children with CP use the same strategies as TD children to adapt to higher gait speeds. 72 gait parameters were compared between groups and between walking speeds by non-parametric analysis (Wilcoxon and Kruskal-Wallis) with a significance level of $p<0.05$.
Results: Gait speed range in CP group (100-140% of CWS) was significantly lower than in TD children (100%-180%). No differences were found between CWS of TD children (0.51±0.05) and S2 in CP (0.56±0.05). A first group of parameters was different

between CP walking at S2, compared to TD children walking at CWS and was also different when the CP group walked at lower speeds. So these deviations could not be related to speed adaptations. These parameters were reduced range of motion (ROM) during push-off (p=0.015), increased sagittal pelvic ROM (p=0.011) and reduced ROM of rectus femoris and hamstrings and peak length of gastrocnemius in children with CP compared to TD children. A second group of parameters did not differ between CP at S2 and TD children at CWS, while they were clearly different when CWS of both groups were compared. So, these gait deviations disappeared in CP group by walking faster: reduced maximal hip extension and hip abduction moment, delayed timing of maximal hip flexion moment and lower peak lengthening velocity of gastrocnemius and hamstrings. A third group of parameters was significantly different between CP at S2 and TD at CWS, while these parameters were not deviant when comparing both groups at CWS. These gait deviations were typical for adaptations of gait pattern at high speed: premature peak knee flexion in stance (p=0.017), increased peak hip flexion in swing (p=0.034), premature maximal ankle dorsiflexion (p=0.044) and extension moment in the hip (p=0.044). These adaptations were comparable in TD and CP children, only premature peak knee flexion seemed to be typically for CP group. Adaptations to higher gait speed (CWS vs. S2 in both groups) further seemed to cause higher power generation in the hip and knee, but not in the ankle. Visual analysis revealed a trend towards a higher peak velocity of gastrocnemius in stance and lower peak velocity of gastrocnemius and hamstrings in swing.

Table 1: Mean±SD of significant kinematics and kinetics

Parameter	CWS TD children	S2 CP
Pelvic tilt	4.18±0.48	9.38±3.77
Hip flexion Swing	41.39±3.41	48.08±5.16
Timing maximal knee flexion	12.33±1.55	7.17±3.07
Timing maximal dorsiflexion ankle	41.58±1.16	22.79±13.27
ROM push off ankle	33.99±3.48	20.00±7.32
Power generation hip stance	0.46±0.19	1.75±0.86
Extension moment hip	1.26±0.29	1.85±0.48
Power absorption hip stance	-0.91±0.33	-1.49±0.44
Power generation hip	1.51±0.17	2.76±1.21
Power absorption ankle	-0.63±0.07	-1.83±1.16
Power generation ankle	5.25±0.04	2.79±1.12

Discussion: Results indicated that correction should be made for gait speed when comparing gait patterns of CP and TD children. Several differences can be found between both groups, but children with CP also frequently use the same strategy as TD children to adapt their gait to higher gait speed. Differences in muscle lengthening velocity, dependent of walking speed, have been documented.[5] Because in this study, no distinction was made between stance and swing phase, which require different muscle actions, this could not be confirmed. Future research should focus on signs of spasticity during gait using EMG data.

References

[1] van der Linden et al. (2002), J Pediatr Orthop 22(6): 800-6; [2] Schwartz et al. (2008), J Biomech 41: 1639-50; [3] Lance (1980); in Spasticity: Disordered Motor Control. Chicago, pp 485-495; [4] Hof AL et al. (1996) Gait Posture (4) 222-223; [5] van der Krogt et al. (2008) ESMAC

Acknowledgements

This study is a part of the IPSA project, supported by IWT Flanders.

The effects of cuff weights on the reduction of limb circumduction and excessive floatation during aquatic gait training

Do Kyeong Lee; Konstantinos Vrongistinos, PhD. & Taeyou Jung, PhD., ATC.
The Center of Achievement, Dept. of Kinesiology, California State University, Northridge

INTRODUCTION

Aquatic therapeutic exercise has been increasingly used for people with neuromuscular disabilities[1). Buoyancy, resistance and drag force can help patients support their body weight, move body segment slowly, and reduce the impact force during aquatic gait[2)3)]. In aquatic gait intervention, cuff weights are often used for people with hemiparesis or hemiplegia in attempt to reduce unwanted floatation of the affected limb. The purpose of this study was to investigate gait responses to a cuff weight (CW) applied on the affected limb during aquatic treadmill walking in people post-stroke.

STATEMENT OF CLINICAL SIGNIFICANCE

The study findings will provide scientific understanding of how individuals with hemipareis can benefit from using a cuff weight on the affected leg during aquatic gait training. These findings will allow clinicians to provide evidence-based aquatic rehabilitation programs for individuals with stroke or similar neuromuscular disorders.

METHODOLOGY

A repeated measures comparative study was used to examine the differences in the gait parameters when a CW was applied, as compared to no CW condition. A total of 22 individuals with a single cerebrovascular accident (11 males/11females; age range 48-88 years, mean 66 years; time post-stoke range 13-264 months, mean 70 months) participated in this study. They were asked to walk on an aquatic treadmill (AquaGaiter™, FERNO, Ohio, 2002) in a chest-depth pool adjusted by a movable floor pool (KBE, Germany, 2002) at a self-selected comfortable speed. Three walking conditions were compared at the same depth water and walking speed: a) no CW, b) CW on the distal shank (right above the ankle), and c) CW on the proximal shank (right below the knee). The underwater 3-dimensional motion analysis system captured all gait variables: spatiotemporal data (cadence, stride time, stride length and foot off time) and lower limb join kinematic parameters (hip, knee and ankle). Six waterproof underwater lenses (Coach Cam, Underwater Camera Co., San Diego, 2005) were placed in water and connected to six digital video recorders on the dry deck. Vicon Motus v.9.2 (Vicon Ltd, Oxford, UK, 2007) was utilized to process gait data. A total of 15 waterproof markers (10mm) were attached to the bony landmarks of the lower extremities using the Helen Hayes marker set (Kadaba, 1990). 5-minute familiarization time prior to each condition and 2-minute rest time between trials were provided for all participants. Each participant was asked to select a comfortable walking speed and preferred CW (between two weights, 0.7kg or 1.5kg) during the familiarization trial. Two 2-minute testing trials were collected for each condition.

RESULT

The affected limb showed significant differences in the sagittal hip kinematics. Significant reductions in the peak hip flexion and extension angles were noted in the affected limb when a CW was used (table 1). The hip sagittal plane excursion was decreased on the affected limb with a CW. No significant differences were found in knee and ankle kinematics. The non-affected side did not show any difference in all gait variables across three training modes. The use of ankle weight showed an increase of the foot off time when compared with No-CW.

Variables	Mean(SD)			P values			
	No-CW (N)	Knee (K)	Ankle(A)	ANOVA	N×K	K×A	A×N
Cadence (steps/min)	54.8(12.6)	54.3(9.93)	53.31(10.08)	0.29	0.604	0.193	0.152
Stride Length (m)	0.37(0.11)	0.37(0.1)	0.38(0.12)	0.085	0.1	0.121	0.766
Stride Time (s)	2.32(0.51)	2.22(0.59)	2.33(0.42)	0.254	0.116	0.108	0.815
Foot off time (% cycle)	58.57(5.94)*	59.52(5.17)	60.23(5.24)*	0.062	0.141	0.243	0.017*
Hip flexion (deg)	35.03(11.37)*	33.14(11)*	32.28(11.8)*	0.003*	0.015*	0.153	0.001*
Hip extension (deg)	-2.63(10.36)*	-2.02(10.03)*	-3.27(10.41)*	0.008*	0.058	0.018*	0.311
Hip abduction (deg)	5.37(8.11)	4.75(7.53)	4.54(7.1)	0.497	0.316	0.654	0.236
Hip adduction (deg)	-4.41(5.27)	-3.88(5.58)	-4.55(5.55)	0.271	0.192	0.178	0.789
Hip internal rotation (deg)	3.71(13.61)	1.9(14.68)	2.8(15.6)	0.414	0.185	0.58	0.518
Hip external rotation (deg)	-25.11(13.94)	-24.55(15.05)	-25.28(13.83)	0.686	0.533	0.481	0.893
Hip flex_ext (deg)	37.66(9.17)*	35.16(9.23)*	35.55(8.38)*	0.01*	0.003*	0.584	0.015*
Hip ab_ad (deg)	9.78(4.53)	8.63(3.82)	9.09(3.75)	0.116	0.04	0.294	0.249
Hip inR_exR (deg)	28.82(10.73)	26.44(11.17)	28.08(10.87)	0.301	0.139	0.36	0.716
Knee flexion (deg)	54.87(15.65)	54.43(16.05)	54.07(15.55)	0.743	0.752	0.802	0.437
Knee extension (deg)	12.43(9.28)	11.07(9.78)	11.17(9.93)	0.142	0.048	0.848	0.093
Knee flex_ext (deg)	42.44(14.47)	43.36(15.2)	42.9(13.86)	0.81	0.518	0.74	0.688
Ankle dorsi flexion (deg)	11.96(12.31)	11.56(12.48)	10.99(11.11)	0.792	0.828	0.712	0.51
Ankle plantar flexion (deg)	-9.6(8.73)	-10.06(9.84)	-8.6(7.88)	0.338	0.667	0.198	0.231
Ankle df_pf (deg)	21.56(12.2)	21.62(14.93)	19.59(8.94)	0.371	0.982	0.399	0.201

Table 1. The affected limb gait data summary. flex_ext: excursion of flexion and extension, ab_ad: excursion of abduction and adduction, inR_exR: excursion of internal rotation and external rotation, df_pf: excursion of dorsi flexion and plantar flexion.

DISCUSSION

The results suggest that placing a cuff weight below the knee or above the ankle can affect the hip kinematics during aquatic walking. The reduction of peak hip flexion at the end of swing phase is considered to be related to a decrease of unwanted floatation of the affected limb. With additional weight on the affect limb, the flexion and extension kinematic values showed similar degrees to those of the non-affected limb. This may contribute to improving kinematic gait symmetry. In addition, an ankle weight was found to help increase the stance time of the affected limb closer to that of the non-affected side, improving stability during stance phase and eventually enhancing temporal gait symmetry. However, the results did not support our assumption on reducing circumduction since we found no kinematic differences in hip abduction/adduction and internal/external rotation. In summary, our findings suggest that cuff weight usage can reduce unwanted floatation of the affected limb and improve gait symmetry, but may not reduce limb circumduction during aquatic walking.

REFERENCES

1) Roesler, H. et al., Gait and Posture; 24: 412-417 (2006)
2) Barela, A. et al., Journal of Electromyography and Kinesiology; 16: 250-256(2006)
3) Miyoshi T. et al., Clinical Biomechanics; 20: 194-201(2005)

IDENTIFYING MOBILITY AND EXERCISE IN DAILY STEP COUNTS

Michael S. Orendurff[1], Jason D. Walker[2]
[1] Movement Science Laboratory, Texas Scottish Rite Hospital for Children, Dallas, Texas;
[2]Dept of Mechanical Engineering, University of Southern California, Los Angeles, California
Michael.Orendurff@tsrh.org

INTRODUCTION

Computerized gait analysis has been used to evaluate the results of interventions on technical gait measures such as knee extensor moments, ankle equinus, hip adduction, or other joint dysfunction in controlled laboratory settings. The improvements observed in these technical measures are a key tool in assessing the outcomes of interventions, but additional data on walking behavior outside the laboratory might also prove valuable. The gait used during activities of daily living (GDL) has been suggested to involve very short duration walking behavior with 40% of all walking bouts lasting <13 steps in a row and 75% of all walking bouts lasting <40 steps in a row before stopping [1]. Improving gait performance on these short duration walking tasks generally used for walking indoors is critically important for mobility for individuals with substantial gait pathology. It has also been shown that very active and very sedentary individuals have similar bout duration distributions [1], suggesting that current analysis methods may be inadequate to detect changes in moderate-intensity long-duration walking behavior. Exercise-level walking behavior also has important health-related benefits if the intensity and duration is sufficient to cause physiological adaptation. Exercise reduces risk factors for cardiovascular disease, cancer, stroke, diabetes, obesity and movement morbidity. For children, exercise-level activity also has a substantial developmental component as most gross motor develop in children is play-based.

CLINICAL SIGNIFICANCE

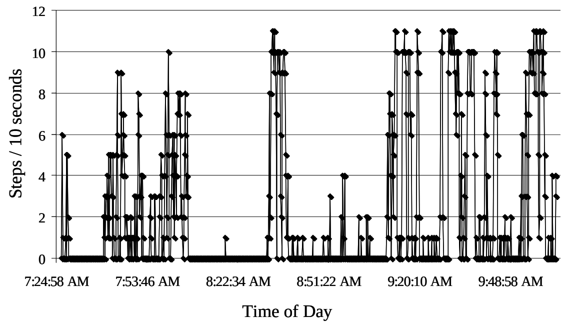
Treatment outcomes may be more fully described by assessing short duration walking used for mobility and long duration walking which may provide an exercise-related health benefit.

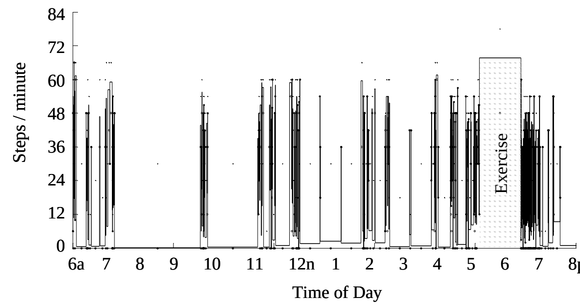
METHODS

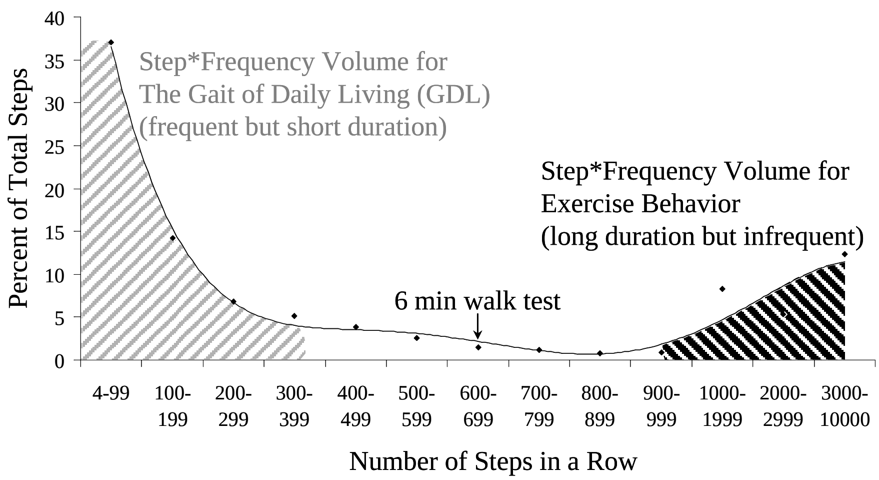
Ten adults, all employed and without known gait disturbances, agreed to participate in this IRB-approved experiment. Subjects wore a StepWatch Activity monitor all waking hours for 14 days (10 weekdays and 4 weekend days). Steps were counted for each 10 second window and written to a file. These data were analyzed using custom code written in MATLAB (The Mathworks, Natick, MA) to determine the duration of moderate intensity exercise behavior for each individual. Activity was judged “exercise” if it met two criteria: prolonged activity with >5 steps in each sequential 10 second period and no break longer than 10% of the current accrued duration. (For example, 10 minutes of moderate activity permitted a 1 minute break—like waiting at a crosswalk before continuing walking.) The output produced a square-wave of activity assessment with the duration, mean intensity, max intensity and minimum intensity within each period of moderate activity and recovery or rest. The goal was to determine when individuals engaged in moderate level exercise behavior.

RESULTS

Raw step count data is shown below, and processed data with exercise bout identified.







The data show that the walking behavior of normal employed adults is frequently of very short duration, and that the vast majority steps taken are during short walking bouts of 4 to 99 steps in a row ($p < 0.001$). The data also identify the infrequent, long duration walking behavior that may be considered exercise. This method may identify an individual's improvement in walking for mobility and their improvement in walking for health benefits or to meet developmental milestones.

DISCUSSION

This is the first attempt to evaluate real-world walking behavior to determine if step count data can be used to identify the frequent, short duration walking behavior used during mobility for the activities of daily living and identify rare, long-duration walking behavior which might be considered exercise. This is important in individuals with orthopedic interventions who want return to exercising or sports after they heal from surgery. It is also important to children whose developmental milestones depend on active play. It may be possible to correlate technical gait measures with GDLs or with exercise behavior to identify critical joint functions which impact these diverse elements of human movement. This initial step * frequency data provides a template for evaluating human gait at the two ends of the spectrum—one for short duration mobility and one for health-benefit exercise duration.

REFERENCE

1. Orendurff, M.S., et al, *J. Rehab Res Dev*, in press.

Partial Body Weight Support Treadmill Training: Clinician's Upper Extremity Muscle Activation During Facilitation of Hemiparetic Limb Movement

Thad W. Buster, Amy J. Goldman, Laura M. Corbridge, Yu Shu, and Judith M. Burnfield

Institute for Rehabilitation Science and Engineering, Madonna Rehabilitation Hospital, Lincoln, NE; E-mail: JBurnfield@madonna.org

INTRODUCTION

Partial body weight support treadmill training (PBWSTT) is commonly used in rehabilitation settings to help individuals recovering from a stroke regain walking ability. The patient wears a body harness which is attached to an overhead suspension system. The suspension system reduces the body weight the patient must support while walking on a treadmill. When hemiparesis is present, a clinician is often required to manually assist with lifting, advancing and stabilizing the hemiparetic leg during each stride. Physical therapists anecdotally report fatigue and transient pain in their arms and trunk during and following PBWSTT sessions. Despite numerous studies highlighting the beneficial impact of this intervention on patient outcomes, the authors are unaware of any published research quantifying the muscular effort exerted by clinicians as they facilitate movement of the hemiparetic limb. Thus, the purpose of this study was to explore biceps brachii muscle activation in a single physical therapist facilitating hemiparetic limb advancement and stability during administration of PBWSTT.

CLINICAL SIGNIFICANCE

The one-year incidence rate of work-related musculoskeletal disorders in physical therapists has been estimated to be 20.7%.[1] Understanding factors that may contribute to injuries to therapists appears warranted to help guide future safe patient-handling policies aimed at maximizing patient outcomes while minimizing clinician injuries.

METHODS

The physical therapist assessed in this study was 35 years old with 9 years of experience working in an inpatient acute rehabilitation stroke program. She had 7 years of experience delivering PBWSTT. Ten adults with hemiparesis secondary to acute unilateral stroke (onset < 1 month prior; mean age = 68±10 years; mean mass = 86±13 kg) were recruited from the same inpatient rehabilitation program to participate in a PBWSTT study.

Each patient completed three PBWSTT sessions prior to formal biomechanical assessment. During the data collection session, the level of body weight support ranged from 15% to 30% across patients and up to 3 clinicians facilitated movement. Each patient's stride characteristics were recorded using footswitches. Surface electrodes were applied to the PT's left and right biceps brachii muscle to document the electromyographic (EMG) pattern during facilitation. Prior to initiating the walking trials, a maximum manual muscle test (MMT) was recorded for the PT's left and right biceps brachii and these data provided a basis for subsequent EMG normalization. The PT facilitated movement of the hemiparetic limb with one arm assisting movement at the knee (KNEE) and the other assisting movement at the foot and ankle (FOOT). The patients' stride characteristics and the PT's biceps brachii activation patterns were recorded simultaneously as the patients walked two minutes at their self-selected free (FREE) and self-selected fast (FAST) walking speeds. These speeds were selected as they are common speeds targeted in the therapeutic setting.

Footswitch data recorded from each patient's hemiparetic limb were used to define each gait cycle (i.e., ipsilateral to the next ipsilateral initial contact). Representative biceps brachii EMG data recorded during 10 hemiparetic limb gait cycles occurring during the final 30 seconds of each walking trial were filtered, rectified and integrated. Data were normalized to the respective left and right biceps brachii MMTs and subsequently expressed as % MMT. A predetermined minimum threshold of 5% MMT identified clinically significant muscle activation. Variables examined included left and right biceps brachii EMG peak, mean and duration. Descriptive statistics and frequency counts characterized the relative effort exerted by the clinician during facilitation. The impact of training speed on clinician muscle activation was compared using Paired T-tests or Signed Rank tests when normality assumptions were violated.

RESULTS (Table 1)

Average walking velocity increased from 48 to 57 m/min between the FREE and FAST conditions in part due to an increase in cadence from 89 to 96 steps/min. Peak EMG: Peak biceps brachii EMG activity ranged from 0% MMT to 106% MMT across patients, walking speeds and facilitation arm. The biceps brachii peak was significantly higher while facilitating the KNEE at the FAST compared to FREE speed ($p=0.037$). Peak activity exceeded 50% MMT in 4 of the 40 recordings, 25% MMT in 14 recordings, and 10% MMT in 29 recordings. *Mean EMG:* The mean biceps brachii EMG ranged from 0% MMT to 33% MMT across conditions. No significant differences were identified between FREE and FAST conditions. Mean activity exceeded 25% MMT in 2 of the 40 recordings and 10% MMT in 19 recordings. *Duration of EMG:* The duration of biceps brachii EMG activity ranged from 0 to 99% across conditions. No significant differences were identified between FREE and FAST conditions. Continuous activity (i.e., >95% gait cycle) was documented in 2 of 40 recordings. The duration exceeded 75% gait cycle in 8 recordings, 50% gait cycle in 18 recordings, and 25% gait cycle in 23 recordings.

Table 1. Summary of clinician's biceps brachii EMG activity during PBWSTT facilitation of the hemiparetic limb at the KNEE and FOOT.

	Peak (% MMT)		Mean (% MMT)		Duration (% Gait Cycle)	
	FREE	FAST	FREE	FAST	FREE	FAST
KNEE Facilitation	20 ± 30	27 ± 32	9 ± 10	10 ± 9	36 ± 36	44 ± 38
FOOT Facilitation	24 ± 18	27 ± 21	11 ± 5	11 ± 6	36 ± 30	41 ± 30

DISCUSSION

Facilitation of hemiparetic limb advancement and stability at a rate of approximately 45 times per minute required substantial clinician muscle effort as evidenced by the four instances of peak EMG exceeding 50% MMT and nearly half of the recordings requiring muscle effort for greater than 50% of the cycle. While this study focused on the therapist's arm muscles, future work focused on muscles of the trunk and shoulders are warranted due to anecdotal reports of pain in these areas.

REFERENCES

1. Campo, M. et al. (2008). *Physical Therapy,* 88(5):608-619.

ACKNOWLEDGEMENTS

Project supported in part by a grant from the Christopher and Dana Reeve Foundation.

MODIFIED CONSTRAINED INDUCED MOVEMENT THERAPY: EFFECTS ON SELECTED GAIT CHARACTERISITICS OF CHILDREN WITH HEMIPLEGIA

Tasos Karakostas [1,2], Coker Patricia [2], Hsiang Simon [3]
[1] Children's Memorial Hospital, Chicago, IL, USA; [2] Medical University of South Carolina, Charleston, SC, USA; [3] North Carolina State University, Raleigh, NC, USA

INTRODUCTION AND CLINICAL SIGNIFICANCE

Constrained-induced movement therapy (CIMT) for the treatment of upper extremity motor deficits of children with hemiplegia has become a popular rehabilitation approach. Improvements have been explained on the grounds of neural plasticity and cortical reorganization. Learning a new activity through extended practice has shown to enlarge cortical representation of healthy muscles [1]. Several patients show new activation in the contralateral motor/premotor cortices after CIMT, while others show increased activation of the ipsilateral motor cortex and supplementary motor area. In general, pediatric CIMT approaches have focused on modifying the traditional CIMT protocols to emphasize improving functional bimanual hand skill or improving fine motor skills in the weaker extremity for tasks typically completed by one's dominant hand [2]. Although there is very limited evidence about the effects of CIMT in the rehabilitation of the lower extremities of adults [3], there is none about the effects of CIMT in the rehabilitation of the lower extremities of children with CP. Some adult studies have successfully modified the typical CIMT protocols in order to use principles of constraint to improve lower extremity function through massed practice and forced use of the affected lower extremity without, however, the use of a restraint on the non-affected upper extremity [4].

Consequently, while there is some limited evidence that the theoretical framework that supports upper extremity-based CIMT could be applied to treatment of lower extremity deficits seen in children with hemiparesis, there is no information regarding how constraint therapy effects the atypical gait patterns of children with hemiplegic CP. If the effectiveness of CIMT on the gait of children with CP were known, then specific protocols could be devised to maximize its potential benefits. Consequently, the purpose of this study was to investigate if participation in a modified CIMT (mCIMT) program will have an effect in the gait of preschool children with hemiplegic CP.

METHODS

Six preschool aged children, all diagnosed with hemiplegic CP, participated in a mCIMT program for six-hours a day, for five consecutive days. Each child wore a restraint on their non-affected extremity while participating in all daily camp activities. The children were engaged in developmentally appropriate activities involving massed practice and intensive training in use of their weaker upper and lower limb. The treatment protocol included shaping of more mature motor skills with repetitive practice. This study was approved by the Institutional Review Board of the institution and informed consent was obtained from the families of all children involved in this study.

Pre and post intervention data on the temporal-spatial aspects of gait were collected at our motion analysis laboratory using the GAITRite walkway (CIR Systems Inc. Clifton, NJ 07012) sampling at 80 Hz. Data were collected with the children walking barefoot and with their AFOs, if they were using any. The testing sequence was randomized. Data were analyzed using a generalized linear model with repeated measures taken on each child.

RESULTS AND DISCUSSION

Heel to heel base of support significantly deceased ($p<.001$) between the affected and unaffected lower extremities directly as the result of the mCIMT, as it did step time differential ($p=.04$) (see Table 1). The results suggest a narrowing of the children's base of support, which may be related to decreased need for positional stability.

Table 1
Parameters that demonstrated a statistically significant difference following the mCIMT intervention. The hemiparetic upper extremity was the right for all subjects.

Parameters	mCIMT	AFO	mCIMT + AFO
Step Time L	0.039	<0.0001	0.6825
H to H Base of Support L	***<0.0001***	0.1509	0.0021
H to H Base of Support R	***0.0003***	0.1145	0.0015
Single Support L (%GC)	0.014	<0.0001	0.0522
Single Support R (%GC)	0.0064	0.0997	0.4308
Double Support L (%GC)	0.0001	<0.0001	0.0403
Double Support R (%GC)	<0.0001	<0.0001	0.0622
Swing L (%GC)	0.0018	0.0824	0.7676
Swing R (%GC)	0.0186	<0.0001	0.0055
Step Extremity Ratio R	0.0071	0.05	0.2191
Step Time Differential	0.0436	0.8747	0.8464

Our results suggest that mCIMT can improve the temporal-spatial aspects of gait, especially those related to base of support and symmetry. Furthermore, it may be that mCIMT could be used for remediation of lower extremity deficits.

REFERENCES

[1] A. Pascual-Leone, "Modulation of motor cortical outputs to the reading hand of Braille readers. ," *Annals of Neurology,* vol. 34, pp. 33-37, 1993.

[2] J. R. Charles, S. L. Wolf, J. A. Schneider, and A. M. Gordon, "Efficacy of a child-friendly form of constraint-induced movement therapy in hemiplegic cerebral palsy: a randomized control trial," *Developmental Medicine & Child Neurology,* vol. 48, pp. 635-42, Aug 2006.

[3] E. Taub, G. Uswatte, and R. Pidikiti, "Constraint-induced movement therapy: a new family of techniques with broad applications to physical rehabilitation-a clinical review," *Journal of Rehabilitation and Research Development*, vol. 36, pp. 237-251, 1999.

[4] S. L. Fritz, A. L. Pittman, A. C. Robinson, S. C. Orton, and E. D. Rivers, "An intense intervention for improving gait, balance, and mobility for individuals with chronic stroke: a pilot study," *Journal of Neurologic Physical Therapy,* vol. 31, pp. 71-6, Jun 2007.

ACKNOWLEDGEMENTS

Medical University of South Carolina: Internal seed grant.

UPPER EXTREMITY MUSCLE FATIGUE THAT INDUCES MUSCLE IMBALANCES DOES NOT INCREASE MOVEMENT INSTABILITY

Deanna H. Gates[1] and Jonathan B. Dingwell[2]

[1]Department of Biomedical Engineering, University of Texas, Austin, TX, USA
[2]Department of Kinesiology & Health Education, University of Texas, Austin, TX, USA
E-mail: jdingwell@mail.utexas.edu Web: http://www.edb.utexas.edu/faculty/dingwell/

INTRODUCTION

Asymmetric motions are common in the workplace [1] and may cause differential repeated loading of joints and muscles. The diverse demands on the muscles surrounding a joint may cause them to fatigue at different rates and to different degrees [2]. Muscle fatigue leads to decreased force production [3]. This could create a force imbalance around the joint, which could lead to abnormal stress distributions [4] within the underlying tissues, resulting in inflammation. Strength imbalances could also lead to movement instability, which could further increase injury risk. The purpose of this study was to determine if local fatigue of the shoulder flexors could generate a muscle strength imbalance about the shoulder and if this would impact the body's ability to maintain stability of the shoulder motion.

CLINICAL SIGNIFICANCE

Fatigue that leads to muscle imbalances and decreased stability could increase risk of injury. It is important to understand how muscle fatigue affects the control of movement stability.

METHODS

20 healthy right-handed (25 ± 2 years) subjects sat in an adjustable chair with seat belts to help them maintain constant posture. They then pushed a low weight (10% of their max. pushing/pulling force) back and forth along a low friction horizontal track in time with a metronome for 5 minutes (Pre-Sawing). Subjects then performed a repetitive lifting task designed to fatigue the shoulder *flexors* for 3 minutes (Fatigue). They then performed the same sawing task for an additional 5 minutes (Post-Sawing). 3-D movements of the arm and trunk were recorded continuously at 60 Hz using VICON. The three rotational angles of the shoulder were calculated using Euler angles. EMG data were collected at 1080 Hz from 9 arm and trunk muscles. Maximum force measurements (MVCs) were taken before and after each task. EMG median power frequency (MdPF) was calculated to quantify muscle fatigue.

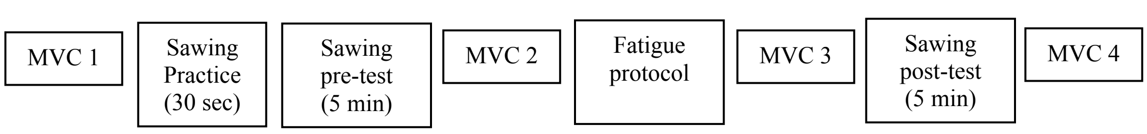


Figure 1. Diagram of the Experimental Procedure

Local dynamic stability [5] was quantified using short-term local divergence exponents (λ_s^*) which indicate the rate of divergence of neighboring trajectories. Positive exponents indicate local instability, with larger exponents indicating greater instability [5]. λ_s^* was calculated for each shoulder angle pre and post-fatigue. Paired t-tests were used to compare λ_s^* values pre and post-fatigue. MVC values were first normalized to % maximum MVC and compared using a 2-factor ANOVA (subjects x trial).

RESULTS

Subject exhibited significant muscle fatigue. Maximal shoulder flexion strength decreased by approximately 20% after completing the fatigue protocol. Strength decreases in shoulder flexion / extension (Fig 2A), and internal / external rotation (Fig 2B) were all significant ($p < 0.006$). The muscle strength also became more slightly unbalanced as a result of fatigue (Fig 2C), but this was not statistically significantly ($p < 0.081$). λ_s* tended to decrease for all shoulder angles (Fig 2D). This decrease was significant for humeral rotation ($p = 0.020$). Thus, humeral rotation became more stable after muscle fatigue.

SUMMARY/CONCLUSIONS

Fatigue led to decreased muscle strength and increased muscle imbalance. Fatigue also led to *decreased* local dynamic instability in the humeral rotation angle. This result suggests that after subjects fatigued, they may have needed to apply greater control to the rotation movements of their humerus. Future work will determine if this increase in stability occurs as a result of increased muscle activity.

REFERENCES

[1] Marras, W.S. et al. (1993) *Spine* 18 (5): 617-28.

[2] Kumar, S. (2001) *Ergonomics* 44(1): 17-47.

[3] Basmajian, J. and DeLuca, C. (1995) *Muscles Alive* 5th Ed.

[4] Gorelick, M. et al. (2003) *Appl Ergon* 34(4): 317-25.

[5] Dingwell, J. and Marin, L. (2006) *J Biomech* 39(3): 444-52.

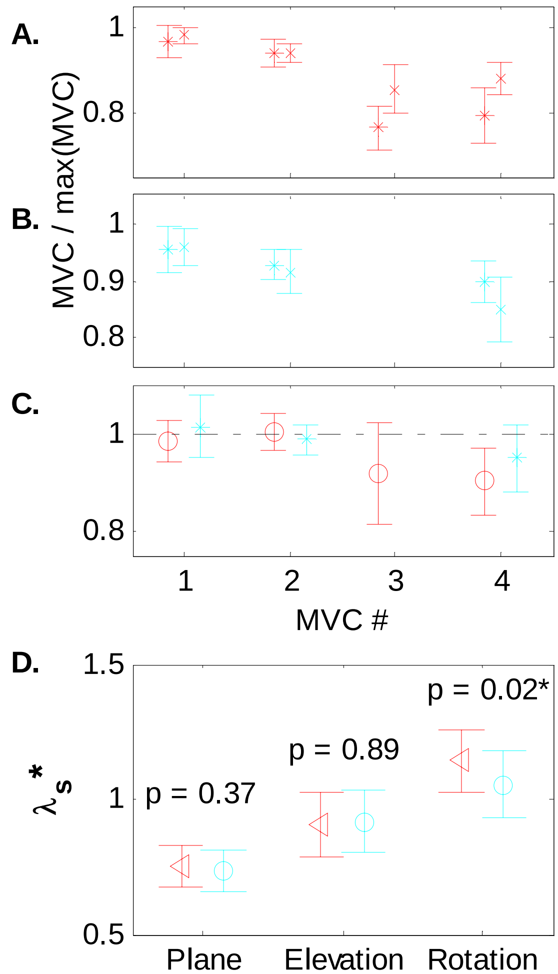


Figure 2. A) Shoulder flexion ('*') and extension ('x') strength decreased with muscle fatigue ($p < 0.001$). **B)** Shoulder internal ('*') and external ('x') rotation strength also decreased with fatigue ($p < 0.006$). **C)** The ratio of flexion to extension strength ('o') and external to internal rotation strength also decreased with fatigue ($p < 0.081$) indicating increased imbalance. **D)** The greatest instability was found in the rotation angle. This instability *decreased* significant after fatigue ($p = 0.02$).

ACKNOWLEDGEMENTS

NIH Grants # 1-R21-EB003425-01A1 and # 1-R03-HD058942-01 (to JBD).

SITTING POSTURAL CONTROL IN DEVELOPMENTALLY DELAYED INFANTS: A STOCHASTIC OR DETERMINISTIC PROCESS?

Anastasia Kyvelidou[1], Regina T. Harbourne[2], Wayne A. Stuberg[2] and Nicholas Stergiou[1,3].
1 HPER Biomechanics Laboratory, University of Nebraska at Omaha, Omaha, NE, USA
2 Munroe-Meyer Institute, University of Nebraska Medical Center, Omaha, NE, USA
3 College of Public Health, University of Nebraska Medical Center, Omaha, NE, USA

Introduction

Researchers have employed the center of pressure (COP) in investigations of postural control during standing in healthy[1] and children with cerebral palsy[2]. Recently, this methodology has also been utilized to investigate sitting postural control in typically developing infants[3,4]. These studies determined that sitting postural control of typically developing infants is a deterministic process that requires the dynamic control of all the involved body segments as well as the sensory systems (visual, vestibular and proprioceptive) that contribute to the achievement of independent sitting[3,4]. However, for a pathologic population such as infants with developmental delays, there is no evidence whether the acquirement of sitting posture involves nonlinear deterministic processes. Therefore, the purpose of this study was to determine whether the development of sitting postural control in infants with developmental delays is a deterministic or stochastic process.

Statement of Clinical Significance

The investigation of the dynamic properties of COP time series of developmentally delayed (DD) infants will support or dismiss the usage of nonlinear analysis of COP time series in this population. Nonlinear analysis of COP time series in typically developing infants during sitting development revealed that nonlinear parameters, such as Approximate Entropy and Lyapunov exponent, were sensitive in capturing subtle changes with development[3]. Therefore, similar analysis of the COP time series in DD infants may reveal the way that these infants manage to control and attain the sitting milestone through different movement solutions, and ultimately explore various therapeutic methods that will enhance the achievement of independent sitting.

Methods

Twenty-seven DD infants that scored 1.5 SD below the mean for their corrected age on the Peabody Gross Motor Scales and who had risk factors for cerebral palsy participated in the study. Infants came to the lab for two sessions per month for four months. For each session (8 in total) three trials of 8.3 sec of unsupported sitting were recorded while COP data were recorded. The COP is comprised of two time series: the anterior/posterior (AP) and medial/lateral (ML) directions. We tested the deterministic properties of the COP signal separately in each direction with the technique of surrogation. The surrogation technique entails the shuffling the original data in order to generate a random equivalent that has the same variance and mean with the original time series. Next, a nonlinear parameter is computed, such as the Approximate Entropy (ApEn) [6], from both the original and the surrogate time series. If the random surrogate produces statistically significant larger values, this indicates the presence of determinism in the original data[5]. All calculations were performed in Matlab. ApEn values of the surrogate data were compared to the ApEn values of the original data using a dependent t-test ($p<0.5$).

Results

Significant differences were found between the original and the surrogate datasets for the resultant COP as well as for each direction of motion, AP and ML ($p<.001$ for resultant COP, $p<.001$ for AP direction, $p<.001$ for ML direction). These results indicated that the fluctuations observed in the original COP time series were clearly distinguishable from linearly autocorrelated Gaussian noise. Thus, the original data were not randomly derived, and therefore, they may be deterministic in nature.

Discussion

These results support the usage of nonlinear analysis of COP time series in infants with DD. Furthermore, they support the presence of a deterministic structure on the COP time series of sitting posture even for infants with DD, suggesting that this is a fundamental feature of human movement variability [7].We believe that further investigation into the structure of the variability of the COP time series is warranted based on our results.

References

1. Riach CL & Hayes KC (1987) *Dev Med Child Neurol* 29,650-8
2. Cherng RJ et al. (1999) *Am J Phys Med Rehabil* 78,336-43
3. Harbourne R & Stergiou N (2003). *Develop Psychobiol* 42, 368-377
4. Bertenthal BI et al. (1997). *J Exp Psychol* 23, 1631-1643
5. Theiler et al. (1992). *Physica D* 58, 77-94
6. Pincus SM & Goldberger AL (1994). *Am J Physiol* 266, H1643–H1656.
7. Stergiou N et al. (2006). *J Neurol Phys Ther* 30,120-9

Acknowledgements: This work was supported by NIH (K25HD047194), NIDRR (H133G040118), the Nebraska Research Initiative, the McDonald fellowship and the Bukey fellowship from the Graduate Studies office at the University of Nebraska Medical Center.

UPPER EXTREMITY MOTION ANALYSIS: CAN THIS BE USEFUL FOR RECORDING AND QUANTIFYING ASPECTS OF MOVEMENT IN CHILDREN WITH CEREBRAL PALSY?

Thorlene Egerton, Teresa Phillips, Robyn Grote
Queensland Children's Gait Laboratory, Brisbane, Queensland, Australia.

Introduction: For many children with cerebral palsy (CP) upper limb dysfunction is a major functional and aesthetic concern. Children can have a range of neuro-motor impairments affecting their movement and contributing to dysfunction. Different impairments respond differently to the treatment options that are currently available (medical, surgical, pharmaceutical and clinical therapies). For clinicians who work with these children, diagnosing the underlying impairments is challenging. In particular routine clinical measures fall short for distinguishing between dystonia and spasticity. This makes intervention planning difficult and outcomes more variable for some children.

Recent developments in our ability to record movement of the upper extremity with motion analysis equipment offer the potential for more sophisticated assessment procedures. In a study by Malfait & Sanger (2007), the kinematics during a reaching task of children with diagnosed dystonia and healthy controls were compared in order to explore the differences in their movement. The reach task involved pointing to two alternate targets as fast as possible. The children with dystonia showed reduced speed, greater variability and pauses at targets. The decreased speed was mostly due to difficulty in reversing direction rather than changes to the flexion/extension velocity profile. However, their method of diagnosing dystonia was not clear and no comparison with children with predominantly spastic problems was made. Sanger et al (2007) also used kinematic analysis in a clinical trial and found that Botox-B injections improved the speed of movement during reaching in children diagnosed with dystonia. Again the method of diagnosis of dystonia was not clear and it may be that the children who showed improvement had a largely spastic component to their movement disorder. Importantly, these studies show the potential of motion analysis as a clinical assessment tool.

This project aimed firstly to develop a method of recording and quantifying some aspects of upper extremity movement. Pilot subjects were tested to develop the protocol and assess the intra and inter subject variability. Secondly, the movement of typically developing children was recorded to establish normative values for the movement variables. The final aim was to record the upper extremity movement of children with CP who could be categorised as predominantly dystonic or predominantly spastic, and to compare them both with the typically developing children.

Clinical Significance: It is hoped that the findings will lead to improvements in clinicians' ability to accurately diagnose movement disorders and monitor changes in quality of movement, and therefore make more effective use of currently available interventions, and optimise outcomes for these children.

Methods: Four typically developing children and four children with cerebral palsy (two with predominantly spastic movement dysfunction and two with predominantly dystonic movement) were recruited for this project. Participants were marked up for 3-dimensional motion analysis (8-camera VICON motion analysis system and Plug-in-Gait upper extremity model). They were asked to carry out up to 20 repetitions of five tasks: drawing a square, hammering a nail, tapping a drum, reaching between a target and their own nose and grasping/releasing a ball. As qualitative aspects of movement are likely to be important in identifying dystonia, data were derived in terms of smoothness (velocity and acceleration data), variability during trajectory, endpoint position variability, inter-joint coordination (angle-angle graphs), length of pauses during reversals of direction, and cycle time at preferred speed. Data from participant groups was graphically presented for the purpose of making comparisons and exploring differences.

Results: The data from the children with predominantly spastic movement problems and data from the children with predominantly dystonic movement problems was compared with the data from the typically developing children.

Discussion: This pilot project has provided objective data of upper limb movement, including aspects of quality of movement, in children with and without cerebral palsy. Larger studies are required and the clinical application of upper extremity motion analysis is yet to be evaluated.

References:

Malfait, N. and T. D. Sanger (2007). "Does dystonia always include co-contraction? A study of unconstrained reaching in children with primary and secondary dystonia." Exp Brain Res **176**(2): 206-16.

Sanger, T. D., S. N. Kukke, et al. (2007). "Botulinum toxin type B improves the speed of reaching in children with cerebral palsy and arm dystonia: an open-label, dose-escalation pilot study." J Child Neurol 22(1): 116-22.

Acknowledgments:

Funding for this project was provided by the Royal Children’s Hospital Foundation, Royal Brisbane Hospital, QLD 4029.

GAIT EVENT DETECTION USING A MULTILAYER NEURAL NETWORK

Adam Miller
Mary Free Bed Rehabilitation Hospital, Grand Rapids, MI USA

INTRODUCTION

Gait data are typically normalized to a gait cycle to facilitate analysis. This requires that the accurate timing of heel-strike and toe-off events be known. Though several methods have been developed to automatically detect events using kinematic data in normal gait, none have been validated for use in pathological gait[1-7]. Consequently, many clinical gait analysis labs use visual inspection of motion capture data to detect gait events. The focus of this paper is to provide a new tool for detecting gait events using kinematic data for all gait patterns.

CLINICAL SIGNIFICANCE

Visual detection of gait events is a laborious process that adds significantly to data processing time. Decreasing or eliminating this time will help gait analysis laboratories operate more efficiently.

METHODS

Relevant motion data were extracted by projecting the heel and toe marker trajectories onto the Y-Z plane (sagittal plane) of the lab coordinate system. The horizontal and vertical components of velocity and acceleration were calculated for each marker. In addition, the foot-floor angle (heel-toe line relative to floor), angular velocity, and angular acceleration were calculated. A sliding window approach was then used to create subsets of normalized time series data for any given frame to be used as neural network inputs. The neural network used in this study was a single hidden layer, feed forward network and was trained using events detected from a cohort of 50 pathologic subjects walking in either the barefoot or shod condition. Patient demographics and gait parameters varied widely in the training dataset. In order to validate the results of the network, events detected using the neural network were compared to events detected using force plates for a total of 40 pathologic subjects equally divided into 2 groups: barefoot and shod. Subject characteristics can be seen in Table 1.

Table 1: Subject Characteristics: mean ± standard deviation

Subject Characteristics	Barefoot (n=20)	Shod (n=20)	Training (n=50)
Gender (M/F)	11 / 9	9 / 11	24 / 26
Age (years)	11.7 ± 3.9	12.5 ± 7.2	13.0 ± 8.6
Height (cm)	140.5 ± 21.8	137.9 ± 23.3	140.2 ± 20.6
Body Mass (kg)	34.6 ± 11.9	39.7 ± 23.9	41.3 ± 19.8
Walking Speed (cm/s)	105 ± 22.3	105 ± 20.5	97.6 ± 20.0
Diagnosis			
CP	14	14	33
TBI	1	1	3
Spina Bifida	1	3	5
Other	4	2	9

RESULTS

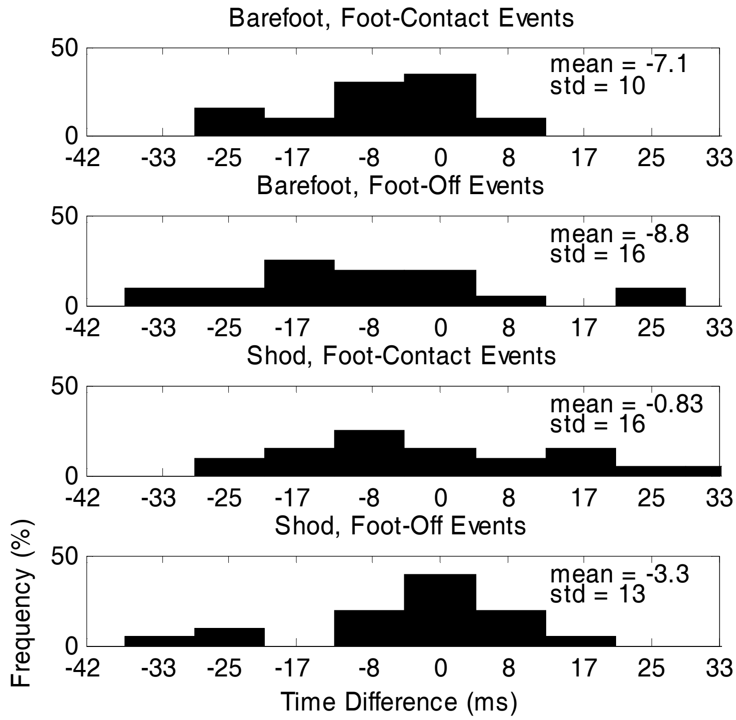


Fig.1. Differences between events detected by the neural network method and the force plate method. A negative difference indicates that the neural network event was detected before the force plate event. Bin widths correspond to 1 frame of data collected at 120 Hz.

In general, the force plate and neural network methods agreed within 25ms (3 data frames) of one another, although in 3 out of 40 foot-off events the difference was 33ms (4 data frames). In the barefoot condition, events detected by the neural network were detected before the force plate an average of 7.1ms for foot-contact events and 8.8ms for foot-off events. In the shod condition, these differences were less than 4ms, or less then one frame.

DISCUSSION

The neural network used in this study was able to accurately and autonomously detect gait events in pathologic gait patterns in both the barefoot and shod conditions. Results from the force plate and neural network methods deviated most when detecting foot-off events in the barefoot condition. The method presented here is a significant improvement over other event detection techniques that are unproven or invalid for use with pathologic gait patterns[1-7]. Once trained, the neural network is completely autonomous and can accommodate any walking speed, gait pattern, brace/shoe combination, or assistive device. This will substantially decrease data processing time for clinical gait labs. In fact, processing time for a single walking trial consisting of 600 frames of data took less than 4 seconds on a computer with a Pentium IV processor.

REFERENCES

[1] Hreljac A, et al. (2000) *J Biomech*;33:783-6.
[2] Zeni JA Jr, et al. (2008) *Gait Posture*;27:710-4.
[3] O'Connor CM, et al. (2007) *Gait Posture*;25:469-74.
[4] Ghoussayni S, et al. (2004) *Gait Posture*;20:266-72.
[5] Mickelborough J, et al. (2000) *Gait Posture*;11:32-7.
[6] Hansen AH, et al. (2002) *J Biomech*;35:135-8.
[7] Stanhope SJ, et al. (1990. *Med Biol Eng Comput*;28:355-60.

THE EFFECTS OF EVERYDAY CONCURRENT TASKS ON OVERGROUND MINIMUM TOE CLEARANCE AND GAIT PARAMETERS

Brian Schulz[1], John Lloyd[1], William Lee[2]
[1] VISN8 Patient Safety Center, James A. Haley VA Hospital, Tampa, FL, USA
[2] University of South Florida College of Engineering, Tampa, FL, USA
E-mail: Brian.Schulz@va.gov   Web: www.patientsafetycenter.com

INTRODUCTION

Minimum toe clearance (MTC) is an indicator of trip risk[1], yet we are unaware of published MTC data for overground gait during concurrent tasks. Here we present a 3D method for determining MTC for all points on the shoe during overground gait in unimpaired subjects walking normally and while performing everyday concurrent motor and cognitive tasks.

CLINICAL SIGNIFICANCE

This is the first study to evaluate the effects on concurrent tasks on MTC. Furthermore, this study also presents an accurate 3D method of determining MTC overground we believe to be an improvement over existing methods. MTC is generally evaluated on a treadmill, which does not take into account the ground height variations, obstacles, and direction changes of overground gait or permit compensatory gait changes to be captured, such as reducing gait speed when performing a concurrent task.

METHODS

Unimpaired men (N=5) and women (N=5) traversed a 3.7m capture volume at their usual walking speed 10 times for each of the following gait conditions (presentation order randomized): 1) normal gait, 2) carrying a laundry basket (concurrent bimanual motor task that occludes vision of feet), 3) answering standardized questions (concurrent executive task), and 4) while carrying a tray with a glass of water on it (concurrent vigilance task and bimanual motor task that occludes vision of feet). Standardized lab shoes in the correct size for each subject and eight integral markers were digitized using a MicroScribe-3DX digitizer (Immersion Corp., San Jose, CA, USA). The floor was digitized using a custom rolling wand. Motion capture data of these markers were used to reconstruct the shoe sole and calculate the minimum shoe-floor clearance for every data frame using custom MATLAB code. The foot was defined to be on the floor when this distance was below zero and the minimum speed of the shoe was below 0.2 m/s. Step length, width, and time were calculated based on average position of heel markers and identified ground contacts. Mixed models in SAS (SAS Institute Inc., Carey, NC, USA) were used for all statistical analysis.

RESULTS

2D MTC on a treadmill has been shown to be non-normal[1]. As these MTC data demonstrate a similar trend, both the mean and median of the MTC data were tested, but this did not alter the results. The results (**Figure 1**) indicate that concurrent tasks during gait significantly effect MTC ($p = 0.01$), maximum toe speed ($p = 0.002$), step length ($p = 0.02$), and stride time ($p = 0.002$). Step width was unaffected by any of the concurrent tasks.

DISCUSSION

MTC differed from previous values in literature (median=9.8 vs. 10.3mm for normal gait[1]), but this may have been due to the different methodology. Subjects appeared to generally

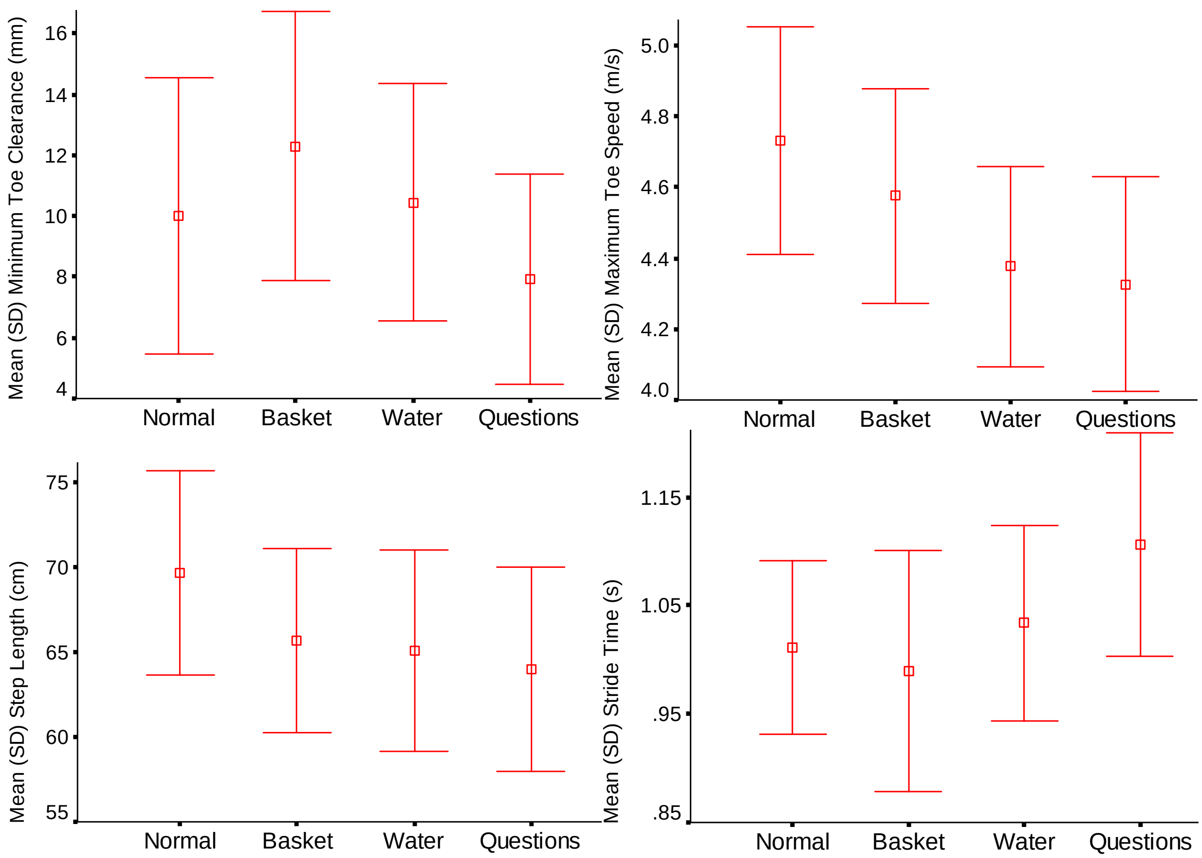


Figure 1. Clockwise from top left: Minimum toe clearance (MTC), maximum toe speed, stride time, and step length by gait type.

compensate for the concurrent tasks examined here by reducing step length, increasing stride time, and decreasing maximum toe speed. However, the effects of MTC appear to be related to the nature of the task performed: The motor-only task (carrying laundry basket) increased toe clearance, the executive-only task (answering questions) decreased toe clearance, and the motor+vigilance task (carrying tray with glass of water on it) had no effect on MTC. It is possible the arousal effects of carrying a greater load (~9 vs. ~1 kg) were responsible for the increased MTC while carrying the laundry basket, but the reduction in MTC for answering questions (an executive task) without a similar reduction for carrying the tray with water glass (a vigilance task) is surprising. It is possible that the effects of the motor and vigilance tasks offset, but further research into the effects of multiple concurrent tasks (i.e. carrying a laundry basket while answering questions vs. just basket or just questions) and of visually occluding the feet would be useful to elucidate the specifics of these effects.

REFERENCES

1. Begg, R., et al. (2007) Gait & Posture, 25(2):191-198

ACKNOWLEDGEMENTS

We acknowledge the support of the Department of Veterans Affairs and the Veterans Health Administration via a Career Development Award and Rehabilitation Research & Development Research Enhancement Award Program (E2964F) and Patient Safety Center of Inquiry (Tampa, FL).

VERBAL FLUENCY TASKS AFFECT GAIT VARIABILITY AND SYMMETRY

Introduction

We investigated the relationship between verbal fluency tasks (letter fluency and category fluency), gait variability and symmetry. The verbal fluency task is a well-defined and standardized neuropsychological test that can be used to explore the effect of attention on gait control during dual-task experimentation. Specifically, category and letter fluency tasks have been used to demonstrate psychological and neurological dissociations between semantic and phonological aspects of word retrieval. Previous neuroimaging and lesion studies have suggested that category fluency (semantic-based word retrieval) is mediated primarily by the temporal cortex, while letter fluency (letter-based word retrieval) is mediated primarily by the frontal cortex [1]. Differential performance in letter versus category fluency tests has been demonstrated in aging-related disorders.

Clinical Significance

Dual-task paradigm might serve to unmask subtle cognitive changes that affect motor variability during gait. Therefore, the combined used of dual-task paradigm and measures of gait variability (using both linear and nonlinear descriptors) may have considerable potential value in identifying patients at risk of falling by monitoring changes in gait patterns, and may lead to effective interventions to prevent falls.

Methods

Twenty-two healthy right-handed young adults walked on a treadmill at their self-selected pace for 3 minutes under three conditions, control (baseline), a letter fluency task, and a category fluency task, while lower extremity kinematics was recorded. Gait variability was calculated from both continuous (joint angles) and discrete variables (joint range of motion) from all strides of each test. The Coefficient of Variation (CoV) was calculated to quantify amount of variability in the discrete variables. The largest Lyapunov Exponent (LyE) was calculated to quantify structure or organization of variability in the continuous variables. A 2×3 (side by condition) repeated-measures ANOVA was performed on COV and LyE for all joints.

Results & Discussion

Our results indicated that gait variability significantly decreased due to both fluency tasks as compared to baseline (Fig. 1). Gait became more periodic and rigid. These results indicated that higher center neural resources are utilized to control gait variability during locomotion regarding both amount and structure. No differences were found between the two fluency tasks indicating that they have similar effect on the control mechanisms of gait variability. Significant symmetry changes were only observed at the ankle and for amount of variability. These results suggested limited effect of the two cognitive tasks on gait symmetry and thus, more involvement of spinal neural control mechanisms for gait symmetry.

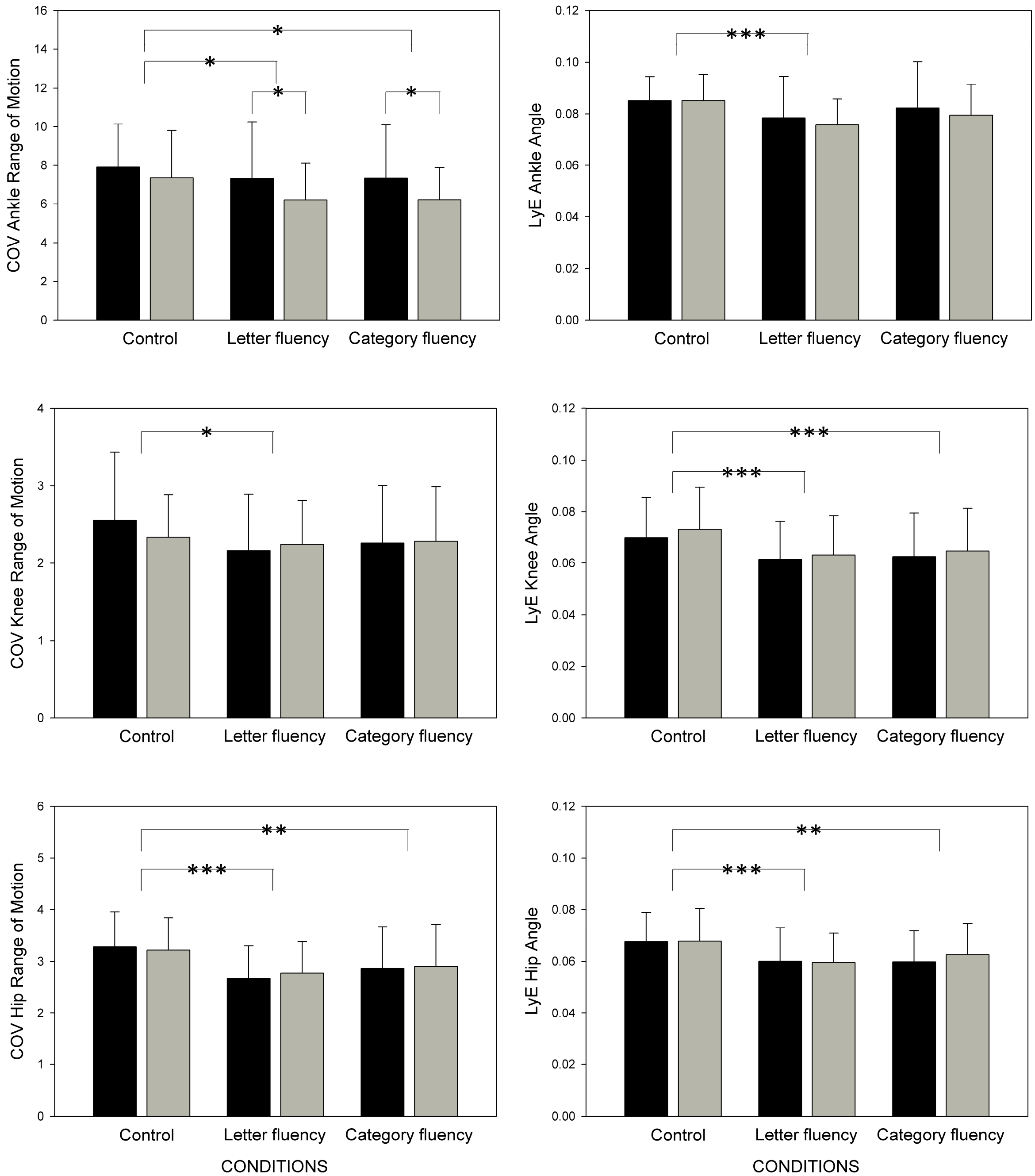


Figure 1. Single-task: walking only; dual-tasks: walking while performing letter and category fluency tasks. *: $p < 0.05$; **: $p < 0.01$; ***: $p < 0.005$. Black bars: right leg, grey bars: left leg.

References

[1] Henri JD, Crawford JR (2004) *Neuropsychology* **18**: 284-95.

Acknowledgements

American Geriatrics Society's Dennis W. Jahnigen Award, Nebraska Research Initiative.

CENTER OF PRESSURE TRENDS DURING 60 S STANDING BALANCE TESTS

Raoul F. Reiser II, Kimberly A. Sisneros, Brian L. Tracy, Caitlin E. Feldman,
Bjørn Jørgensen, and Leah N. Hitchcock
Department of Health and Exercise Science, Colorado State University, Ft. Collins, CO
Email: rfreiser@cahs.colostate.edu

INTRODUCTION

Normal human aging results in neuromuscular decline that increases the likelihood of falls [1,2]. Assessment of center of pressure (COP) motion under the feet during quiet stance is a common way to assess balance and the accompanying risk for falls. Typical trial durations range from 20 to 80 s [3]. It is generally accepted that longer trial durations will produce results with greater magnitudes of sway and other variables due to the increased chance for a random event to occur that compromises balance. However, the possibility exists that these events are not random, but instead could be associated with settling during the initial stages of the trial and/or unsettling related to boredom, loss of attention to the task, and fatigue later in the trial. The purpose of this investigation was to determine if consistent time related trends exist for typical anterior/posterior (A/P) and medial lateral (M/L) measures of balance.

CLINICAL SIGNIFICANCE

A better understanding of how sway parameters adjust/trend during a relatively long trial will help determine appropriate trial durations and guide the interpretation of results.

METHODS

19 young adults (18-30 yrs), 19 healthy elderly with no history of falls (65-90 yrs), and 13 elderly who had fallen at least once in the last year with no apparent cause (65-90 yrs), participated in this study. Ground reaction forces under each foot were sampled at 100 Hz and low-pass filtered at 10 Hz with two synchronized Bertec 4060-10 force platforms mounted side-by-side. Individual force plate recordings were combined, calculating the net COP measures assessed in this investigation (A/P and M/L sway, path length, maximum velocity, standard deviation, and elliptical area).

Since sway in the M/L direction has been shown to be a predictor of future risk for falls [4], postural sway testing with the feet together may be more sensitive when discerning fallers from non-fallers [5]. As such, subjects were asked to stand with one foot completely on each platform with the feet as close together as possible without touching. Subjects stood erect with arms at their sides in a comfortable position. Eight trials (four eyes-open, four eyes-closed) were conducted in a random order. Subjects stood for ~10 s before 62 s of data were sampled while standing as still as possible. ~2 min rest was given between trials. After cropping the first and last second, the middle 60 s and each successive 20 s "bin" were analyzed independently. A/P variables were normalized to base of support length and M/L to width. Results were averaged to produce a representative value for vision and no vision conditions. Since the 60 s path length would represent the sum of the path lengths from each bin, this variable was divided by 3 to make comparable to the 20 s bins. 3 x 4 (subject group x 20 s bins and entire 60 s trial) repeated measures ANOVAs were performed with Bonferroni adjustment to account for the relatively large number of variables (n_v = 9 within each vision condition). Significance was assessed at $p < 0.05/n_v = 0.006$. Main effects for duration were analyzed with four factor repeated measures ANOVAs with Bonferroni post-hoc analyses.

Results

A significant group by duration interaction existed for A/P maximum velocity with vision where the young adults exhibiting a decrease (settling) from the first to second bin and no further decrease into the third bin, the elderly non-fallers exhibited gradual settling across the three bins, and the elderly fallers exhibited no settling until the third bin. Though not significant, interactions approached $p < 0.006$ for A/P path length with vision and sway and standard deviation without vision ($0.015 \geq p \geq 0.013$). Trends in these variables resemble those of the A/P maximum velocity with vision for each group, but not to the same extent.

Regardless of significant interaction, all variables contained significant main effects for duration. When comparing the overall 60 s trial to individual 20 s bins, all variables except path length were greater when computed over the entire trial. Path length tended to decrease across each successive bin with the 60 s value (divided by 3) similar to the middle bin. A significant settling effect (decrease in value across one or more successive bins) was observed in all variables except A/P standard deviation without vision and M/L standard deviation with vision. Several variables exhibited a possible unsettling effect (increase in value in the last bin compared to the middle bin), to include A/P and M/L sway and standard deviation as well as elliptical area with vision. However, unsettling was not statistically significant.

Discussion

Our results demonstrate that even though random events are more likely to occur with longer trial durations, as indicated by the majority of variables being greater when computed over 60 s as compared to a single 20 s bin, there are relatively consistent trends over the course of 60 s. A period of settling exists for most variables beyond the ~10 s given prior to measurement. The time it takes to settle appears to be affected by both age and risk for falling. It is also possible that some unsettling occurs near the end of a 60 s trial. For the most stable results it may be appropriate to assess the last 20 s of a 40 s trial in young adults and the last 20 s of a 60 s trial for elderly adults. However, there appears to be important information related to fall risk in the time it takes to settle and stabilize during a standing balance test, with additional time needed for those at risk of falling. While further exploration of this phenomenon is necessary, it may improve the power of this test to screen for those at risk above and beyond current assessment of just the relative magnitudes of each variable.

References

1. Nevitt, Cummings et al. *J Am Med Assoc*, 261: 2663-2668, 1989.
2. Robbins, Rubenstein et al. *Archives Internal Med*, 149: 1628-1633, 1989.
3. Mizrahi, in *Clinical Biomechanics* (Dvir, editor). Harcourt, Inc: Philadelphia, 2000.
4. Maki, Holliday et al. *J. Gerontology: Med Sci*, 49:M72-M84, 1994.
5. Melzer, Benjuya et al. *Age Ageing,* 33: 602-607, 2004.

Acknowledgements

This publication was supported by a sub-award of CDC Grant R49/CCR811509 through the Colorado Injury Control Research Center to RFR and BLT and NIH Grant K01/AG19171 to BLT. Its contents are solely the responsibility of the authors. KAS was supported by the McNair Scholars Program.

OBESITY DOES NOT INCREASE EXTERNAL MECHANICAL WORK DURING WALKING

Raymond C. Browning[1], Craig P. McGowan[2], and Rodger Kram[3]
[1]Department of Health and Exercise Science, Colorado State University, Ft. Collins, CO
Email: browning@cahs.colostate.edu
[2]Department of Mechanical Engineering, University of Texas, Austin, TX
[3]Department of Integrative Physiology, University of Colorado, Boulder, CO

INTRODUCTION

The prevalence of obesity continues to increase and is due, in part, to a lack of physical activity. Walking is the most common form of physical activity and the net metabolic rate during walking (C_w, Watts/kg) is ~10% greater in obese vs. normal weight adults [1]. There are several determinants of C_w that could contribute to the greater C_w in obese adults. These include external mechanical work (W_{ext}), body weight support, sagittal leg swing, lateral leg swing circumduction and balance. In normal weight adults, W_{ext} accounts for ~50% of C_w [2]. To date, the effects of obesity on W_{ext} have not been reported. The purpose of this study was to compare W_{ext} (individual limbs method [3]) between obese and normal weight adults across several walking speeds. We hypothesized that W_{ext} (J/step) would be greater in obese adults but W_{ext} normalized to body mass would be similar in obese and normal weight adults.

CLINICAL SIGNIFICANCE

A better understanding of the determinants of the greater metabolic cost of walking in obese adults may aid in developing effective physical activity interventions to prevent and treat obesity.

METHODS

Twenty adult volunteers, 10 obese, mass = 111 (20.5) kg, BMI = 35.6 (5.0) kg/m^2 and 10 normal weight, mass = 68 (10.2) kg, BMI = 22.1 (1.6) kg/m^2, (mean (S.D.)), participated in this study. Each subject performed six level walking trials on a dual-belt force measuring treadmill. The treadmill speeds were 0.50, 0.75, 1.00, 1.25, 1.50, and 1.75 m/s and the trial order was randomized for each subject. We collected right leg vertical, anterior-posterior (AP) and medio-lateral (ML) ground reaction forces (GRF) and moments about the AP and ML axis for 10 s at 1000 Hz and low-pass filtered the data at 12 Hz. We used the individual limb method (ILM) to calculate external work done on the center of mass [3] during five consecutive strides. The average force data for the right leg were phase-shifted by 50%, assuming symmetry, to emulate forces produced by the left leg. CoM velocity was determined by integrating the combined forces from both legs acting on the center of mass (CoM) [4]. Individual limb power was calculated from CoM velocity and right leg GRF and ILM work was calculated by integrating the power curve with respect to time. The timing of double and single support within a step cycle was determined from the phase-shifted vertical GRF data. A two-factor (obesity and speed) ANOVA with repeated measures determined how obesity and walking speed affected W_{ext}. A criterion of $p<0.05$ defined significance.

RESULTS

Absolute W_{ext} (J) per step was greater in obese vs. normal weight adults at each walking speed, but as hypothesized, relative W_{ext} (J/kg) was similar between the groups (Figure 1). Step frequency was not different between the groups. Normalized W_{ext} during double support was also similar between the groups. During double support, the trailing leg performed the vast majority of the positive work and the leading leg performed the vast majority of the negative work. Wext was affected by walking speed but there were no significant group (obese vs. normal weight) differences.

DISCUSSION

Our results demonstrate that the W_{ext} performed during walking was not affected by obesity. The increase in W_{ext} at faster walking speeds was due to an increase in both double and single support work. The obese subjects had greater ML GRF's, greater ML CoM velocities, and performed more lateral W_{ext}. However, the contribution of lateral W_{ext} to the total work done per step was still relatively small (~3% and 5% for normal weight and obese at 1.25 m/s, respectively). These data suggest that W_{ext} is not responsible for the greater C_w of walking and that moderately obese adults do not adjust their gait to conserve mechanical energy. Future studies that examine the role of leg swing and lateral stability on the metabolic cost of walking in obese adults are needed.

REFERENCES

1. Browning, R.C., et al., J Appl Physiol, 2006. **100**: p. 390-398.
2. Grabowski, A., C.T. Farley, and R. Kram, J Appl Physiol, 2005. **98**: p. 579-583.
3. Donelan, J.M., R. Kram, and A.D. Kuo, J Biomech, 2002. **35**(1): p. 117-24.
4. Cavagna, G.A., J Appl Physiol, 1975. **39**(1): p. 174-179.

ACKNOWLEDGEMENTS

NIH M01-RR00051 supported this research.

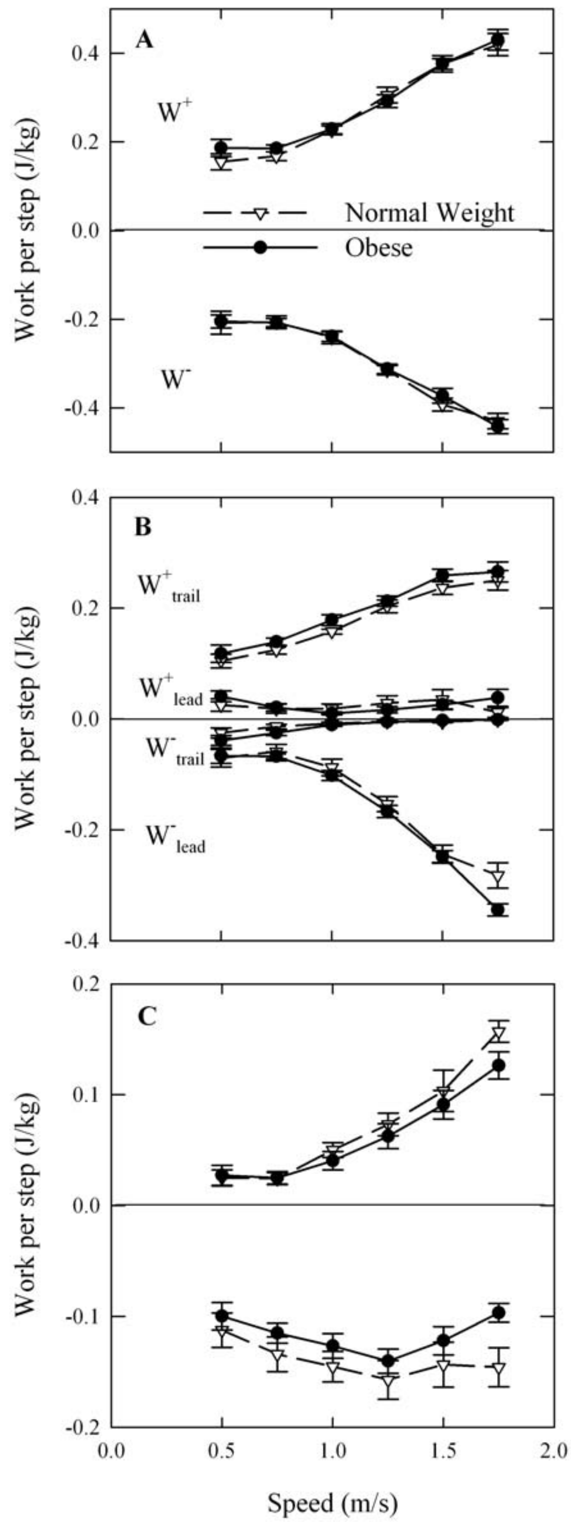


Figure 1. Mean (SE) normalized W_{ext} during a step (A), double support (B) and single support (C) for obese (circles, solid line) and normal weight (triangle, dashed line) adults. W_{trail} is the work done by the trailing leg and W_{lead} is the work done by the leading leg.

A BIARTICULAR EMG-ASSISTED MODEL
TO QUANTIFY CO-CONTRACTION AT THE KNEE AND ANKLE DURING WALKING

Tasos Karakostas [1], Hsiang Simon [2], Mohamad Parnianpour [3], Necip Berme [4]
[1] Children's Memorial Hospital, Chicago, IL, USA
[2] North Carolina State University, Raleigh, USA
[3] Sharif University of Technology, Tehran, Iran
[4] The Ohio State University, Columbus, USA

INTRODUCTION AND CLINICAL SIGNIFICANCE

The development of accurate, non-invasive methods to calculate individual force-time histories during movement across multiple joints is one of the greatest challenges in the study of human motion (White & Winter, 1993). In addition, the ability to precisely define muscle balance across a joint is of primary importance to clinicians who manage neurological or orthopedic conditions, such as cerebral palsy or anterior cruciate ligament deficient knees.

It was the purpose, therefore, of this study to introduce an EMG-assisted model to quantify co-contraction, i.e., muscle balance and imbalance among selected musculotendon units comprising the synergistic and antagonistic muscles involved in ankle plantar/dorsi flexion and knee flexion/extension during normal gait.

METHODS

Description of the mechanical response of the muscle model was based on previous original work (Hill, 1938; Zajac, 1989), but incorporated individual muscle length, velocity, and excitation considerations for muscle contractions. Processed EMG represented the neural input to the muscle. A musculoskeletal model defining joint kinematics, and line of action and architecture of the musculotendon units of the left lower limb, was developed by modifying a previously introduced model (Delp et al., 1990b). Muscle kinematics were calculated in conjunction with three dimensional motion capture and analysis. Individual muscle force as a function of length and level of excitation was also inquired as input to the model, and was established from a series of isokinetic calibration contractions and computer simulations. Co-contraction was measured as an index (CCI) using:

$$\frac{\sum F^M_{Total}}{\sum F^M_{Agonists}} - 1 = CCI$$

where $\sum F^M_{Total}$ is the total muscle force at the joint, and $\sum F^M_{Agonists}$ is the total muscle force of the agonist muscle groups at the joint. The output of the model was validated using the model-predicted and inverse dynamics-measured joint moments.

Four subjects underwent instrumented gait analysis to compute joint kinematics and kinetics. Electromyographyic information (EMG) was collected for the triceps surae, peroneous brevis, tibialis anterior, extensor digitorum, rectus femoris, vastus lateralis, vastus medialis, biceps femoris long and short heads, and semitendinosis/semiembranosis. The walking EMG signals were processed and normalized using activity levels collected from isokinetic calibration contractions.

RESULTS AND DISCUSSION

Results indicate that the moment curves, predicted and measured, matched closely in shape (see Figure 1 for an example). The correlation between the gait moments derived from the two approaches ranged from 0.78 to 0.97 and from 0.73 and 0.91 for ankle and knee joints respectively. The root mean square (RMS) difference between moment curves over one walking stride ranged between 5.3 N.m and 22.2 N.m. The results of this study were, in general, similar or better than those previously reported (White & Winter, 1993). The timing of co-contraction is in agreement with previous studies (Falconer & Winter, 1985). However, to the best of our knowledge, no other studies in the past have presented the quantification of co-contraction at the ankle and knee during walking gait by implementing the estimation of an index, which might allow comparisons. Thus, from a functional perspective the results of the model suggest that co-contraction at the ankle and knee joints during gait occurs at the time when stability is required.

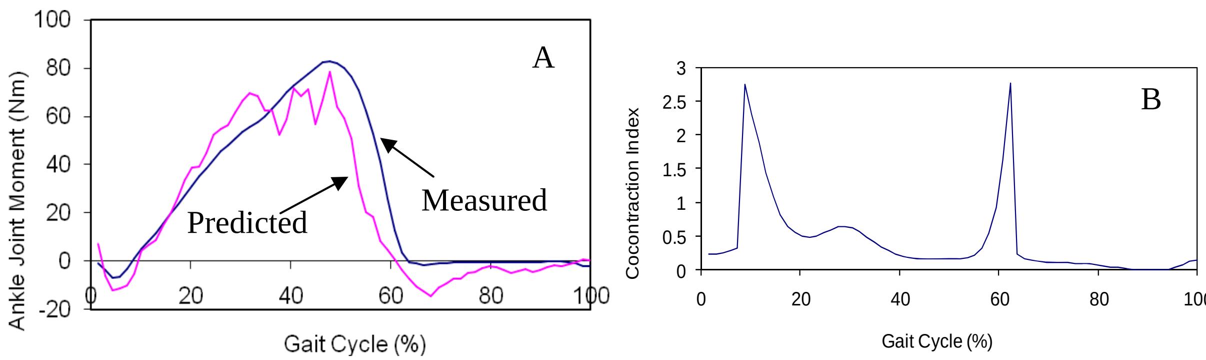


Figure 1. Comparison between the ankle joint moment predicted by the EMG-assisted model and the one measured from the inverse dynamics (A); Co-contraction index for the ankle joint (B).

While this study is limited by the number of subjects, the nature of the moment curve differences suggests that the present EMG-assisted model is essentially correct. However, there is room for improvement. For example, the temporal inconsistencies between the measured and predicted moments can be attributed to a variety of factors, such as the placement of the electrodes or the contribution of the passive structures at the joint.

REFERENCES

Delp, S.L. et al. (1990b). *IEEE Transactions in Biomedical Engineering*, **37**, 757-67.
Falconer, K., & Winter, D. A. (1985). *Electromyography and Clinical Neurophysiology*, **25**, 135-49.
Hill, A. V. (1938). *Proceeding of the Royal Society of Biology, London.*
White, S.C., Winter, D.A. (1993). *J. of Electromyography and Kinesiology*, **2**, 217-31.
Zajac, F.E. (1989). *CRC Critical Reviews in Biomedical Engineering*, **17**, 359-411.

ACKNOWLEDGEMENTS

Kevin Granata
NIDRR (grant #H133E30009).

Influence of Initial Stance Configuration on Gait Initiation

Elan Dalton, Doug Doidge, Shinichi Amano, and Chris J. Hass

Department of Applied Physiology and Kinesiology, University of Florida, Gainesville, FL

INTRODUCTION

Gait initiation (GI) is the transition from quiet stance to steady state locomotion. GI is characterized by initial postural adjustments that result in a center of pressure (CoP) shift in the posterior direction and towards the swing limb (Brunt 1999). The CoP displacement prepares the body for single limb support by shifting the center of mass (CoM) towards the stance limb while simultaneously generating momentum in the anterior direction. These adjustments, which precede foot off, represent the postural phase of GI that enable the subsequent anteroposterior (AP) oriented propulsive phase.

CLINICAL SIGNIFICANCE

Previous reports suggest that older adults and patient populations are inefficient at generating whole-body momentum during GI. The observed difficulties producing the initial mediolateral (ML) weight shift corresponds to a consequent slowness in initiating gait. If an adjustment could be made to assist with the self-initiated postural destabilization while still maintaining the propulsive outcome, GI disability or inefficiency in older adults might be reduced. While the effects of an external lateral assist (Mille 2007) and the influence of modifications to the initial stance width have been evaluated (Rocci 2006), modifications to foot placement in the AP direction have not been investigated.

METHODS

Ten healthy adult subjects (24.8 ± 4.4 yrs, 1.7 ± 0.1 m, 71.0 ± 12.1 kg) initiated gait under three conditions to determine whether manipulations to initial stance configurations would have beneficial consequences. A ten camera Vicon motion analysis system (120 Hz; Vicon Peak, Oxford UK) and a single Bertec force platform (1200 Hz; Model 4060, Columbus, Ohio) were used to collect data. Subjects stood with a "comfortable" foot placement and self-selected their initial swing limb (limb taking the initial step). Initial stance positions were marked and used for the baseline configuration (normal condition) and as a reference point for staggered trials. Modifications to the stance condition were obtained by positioning the swing limb half a foot length anteriorly (forward condition) and half a foot length posteriorly (behind condition), while maintaining stance width. Participants performed 10 trials of gait initiation for each condition. The difference between quiet standing and maximum vertical force under the swing limb, representing the mediolateral weight shift, the peak propulsive force under the stance limb, and whole-body CoM velocity were analyzed. A repeated measures ANOVA was used to test for group stance condition differences with statistical significance at a $p \leq 0.05$.

RESULTS

The amount of swing limb loading during the postural phase was greatest in the behind condition (***p***=.018 behind to normal, ***p***=.001 behind to forward) and least in the forward condition (***p***=.006 normal to forward) (behind: 20.8% body weight (BW), normal: 17.1% BW, forward: 12.6% BW). The magnitude of the propulsive force under the stance limb was significantly greater for the behind compared to the forward conditions (***p***=.001), whereas no significant differences were observed between the normal and forward conditions (behind: 28.3 % BW, normal: 24.4% BW, forward: 24.7% BW). Whole body CoM velocity at the end of the second step did not significantly differ across stance conditions (behind: 1.26 m/s, normal: 1.28 m/s, forward: 1.23 m/s) indicating that gait velocity was maintained across conditions.

DISCUSSION

Elderly and patient populations exhibit delays in gait initiation with altered anticipatory postural adjustments for the lateral weight shift. These altered characteristics may reflect impaired interactions between posture and locomotion as difficulties in ML weight shifting leads to a reduced propulsive force and slower GI. The current study demonstrates that moving the swing foot forward in stance could potentially reduce the demands on the mediolateral weight shift mechanism, while maintaining the forward movement outcome of gait initiation. Thus, manipulating the stance position may affect the interaction between posture and locomotion which may have therapeutic potential for improving movement function in older and/or populations with disabilities. These findings in young healthy population suggest that this technique warrants further investigation in individuals with GI difficulties.

REFERENCES

Brunt D, Liu SM, Trimble M, Bauer J, Short M. Principles underlying the organization of movement initiation from quiet stance. *Gait & Posture* 1999;10(2):121-128

Mille ML, Johnson Hilliard M, Martinez KM, Simuni T, Rogers MW. Acute effects of a lateral postural assist on voluntary step initiation in patients with Parkinson's disease. *Movement Disorders* 2007; 22(1): 20-27

Rocchi L, Chiari L, Mancini M, Carlson-Kuhta P, Gross A, Horak FB. Step initiation in Parkinson's disease: Influence of initial stance conditions. *Neuroscience Letters* 2006;406(1-2): 128-132

ACKNOWLEDGEMENTS

The authors would like to thank Dr. Joe Nocera & Dr. Kim Fournier for their assistance throughout the project.

Gait Performance in Typically Developed Children and Children with Cerebral Palsy (CP)

Sarah Rosen BSc.[1,2], Carole A. Tucker Ph.D[3], Rahamim Seliktar Ph.D[1]
[1]School of Biomedical Engineering, Science and Health Systems, Drexel University, Philadelphia, PA, USA
[2]Shirners Hospital for Children, Philadelphia, PA, USA
[3]Department of Physical Therapy, Temple University, Philadelphia, PA, USA

INTRODUCTION

Existing work in the field of 3D motion analysis rely on the use of kinematic and kinetic data for inverse dynamics calculation of net joint moments and powers during gait. It has been commonly accepted to assume repeatability of the gait event "gait cycle" for the purpose of tracking differences between individuals and changes in performance due to intervention; yet little has been done to validate the "cyclicity" assumption. Seliktar et al. in 1978 [1] developed a "gait consistency test" based on the force time integral (the impulse) of the Ground Reaction Forces. The test was based on the inference that the kinematic repeatability requires that the impulse of the ground forces over a complete cycle be zero, yielding equal velocity of the body center of mass at equivalent positions in consecutive cycles.

$$\int Fdt = \Delta mv \rightarrow if \ \ \Delta v = 0 \ \ than \int Fdt = 0 \quad \text{(Eq. 1)}$$

The reported results of the study were obtained on 28 adults with normal and pathological gait, and consistency was defined as the sum of the A-P impulse over one full gait cycle divided by the absolute values of impulse for both the braking and propulsion phases.
In the current study, the above approach was adopted and applied to three groups of children populations: typically developed children, children with diplegic CP and children with hemiplegic CP, to see whether this consistency test could be applied to typical and pathological gait of children.

CLINICAL SIGNIFICANCE

The longer term objective of the present work is to provide an analytical means to identify quality of gait analysis data based on ground reaction forces with the aim of improving subsequent interpretation of gait analysis.

METHODS

Subjects

Retrospective data of a sample of 53 ambulatory children who underwent gait analysis testing at Shriners Hospital for Children (SHC) - Philadelphia Motion Analysis Laboratory was used in this analysis. The study sample included three groups of children. The first group included children with typical development, 8 males and 8 females, ranged from 7-17 years of age (Mean=11.2, SD= ±2.1). The second group consisted of children with diplegic CP, 8 males and 8 females, ranged from 9-17 years of age (Mean=12.5, SD= ±2.0). The third group included children with hemiplegic CP, with either left or right sides affected, 10 males and 11 females, ranged from 10-17 years of age (Mean=12.9, SD= ±2.3). The analysis included 2 walking trials for each subject and an average of 2 gait cycles per trial.

Data Collection

The data used in this study were collected during gait analysis utilizing motion capture and force plate data collection during level ground ambulation. The Motion Analysis Laboratory at SHC–Philadelphia is equipped with an 8-camera MX-Vicon motion capture system (Vicon Motion systems, Lake Forest, CA) and 4 AMTI force plates. Analog force plate data were synchronized and collected through the Vicon system software.

Data Analysis

Force plate data in the anterior-posterior plane, collected at 1200Hz, was extracted from the Vicon's c3d files and imported into Matlab (The Mathworks Inc. 6.5, Natick MD) for further analysis. A custom-written Matlab program was used to calculate gait consistency, as defined in Seliktar et al. [1].

RESULTS

Our results for a pediatric population demonstrate that children can consistently ambulate within a motion analysis laboratory setting. The results were extremely close to the findings of Seliktar et al. [1] and show consistency is achieved with a ratio below 6.5%. Quantitative results are presented in Table 1.

Population	Average Consistency value [%]
Children with Typical Development	3.44 ± 2.48
Children with Diplegic CP	3.59 ± 2.47
Children with Hemiplegic CP	4.09 ± 2.58
Entire population with no regards to condition	3.77 ± 2.56

Table 1: Results of the consistency test for typically developed children and children with diplegic and hemiplegic CP.

DISCUSSION

The results of the study support the findings of Seliktar [1] and our pilot study defines an inconsistent gait in children to be one in which the consistency value is above 6.5%. The three populations' values are very close and support the assumption that consistency values are similar for children with typical development and children with different types of cerebral palsy. This measure can provide a tool to assess whether a specific gait cycle, chosen for analysis is in fact a representative gait cycle of the child.

ACKNOWLEDGMENT

The authors would like to thank the staff at Shriners Hospital for Children - Philadelphia Motion Analysis Laboratory for its support of this study and the support of the Calhoun Fellowship Program of the School of Biomedical Engineering Science and Health Systems at Drexel University.

REFERENCES

[1] Seliktar R, Yekutiel M, Bar A., Gait consistency test based on the impulse-momentum theorem, Prosthetics and Orthotics International, 3, pp. 91-98, 1979.

CONVENTIONAL GAIT ANALYSIS IN ELDERLY

Suputtitada Areerat, Professor Dr.[1], Saguanrungsirikul Sompol Associate Professor Dr.[2]

[1] Department of Rehabilitation Medicine, [2] Department of Physiology,
Faculty of Medicine, Chulalongkorn University, Bangkok, Thailand

INTRODUCTION

Spatial and temporal gait measurement are useful for follow up the treatments in various conditions such as spasticity, dystonia, movement disorders, pain, musculoskeletal problems, etc. Conventional analysis can be used as the cost-effective method in the community gait analysis or in the clinical setting that would like to evaluate only simple spatial and temporal gait measurement. Pre-and post-treatment gait analysis need to use the same method; either computerized or conventional analysis. Since the computerized gait analysis machines are quite expensive that not available in most hospitals. The objective of this study was to measure spatial and temporal descriptors by conventional gait analysis. The related factors of gait were also measured.

PATIENTS/MATERIALS and METHODS

Spatial and temporal gait measurement on paper with ink as the conventional gait analysis were measured. The related factors of gait were also measured. Both were report as descriptive data. Comparison between women and men were done with student unpaired t-test, significant at $p<0.05$.

RESULTS

Participant included 152 normal elderly (age 71.07 $\pm$ 4.87 yrs); 100 women (age 70.44 $\pm$ 4.00 yrs, BMI 24.03 $\pm$ 3.27 kg/m^2), 52 men (age 72.27 $\pm$ 6.06 yrs, BMI 24.37 $\pm$ 2.99 kg/m^2). There were no statistically significant different in all spatial and temporal gait measurement; stride length, step length, step width ,foot angle, cadence, step rate, stride time, step time and walking speed between women and men. The mean stride length was 72.05 $\pm$ 17.03 cm, step length was 36.08 $\pm$ 8.69 cm, step width was 9.93 $\pm$ 3.67 cm, foot angle was 8.41 $\pm$ 4.03 degree , cadence was 93.92 $\pm$ 26.65 step per minutes , and walking speed was 4.57 $\pm$ 1.54 metres per minute. Q-angle (stand) was 17.91 $\pm$ 4.02 degree , Q-angle (supine) was 17.44 $\pm$ 4.15 degree, leg length discrepancy was 0.14 $\pm$ 0.35 cm, popliteal angle was 26.58 $\pm$ 14.77 degree, knee varus was 11.2%, knee valgus was 84.9 %, heel cord angle was 4.22 $\pm$ 1.75 degree**.**

DISCUSSION

In the clinical setting that would like to evaluate only spatial and temporal gait measurement, not included kinematics (limb motion assessed by the use of motion analysis system and markers across joints),electromyography and kinetic (forces exerted by body using in-ground force plate tranducers) , the conventional analysis is cost effectiveness. The time used was only 10 minute in each case. Since the inconvenience of foot placement in the paper with ink that make sticky of both feet, this may interfere normal gait pattern. In some clinical setting still use this conventional method to analyse only simple spatial and temporal gait measurement. The normative data in eldery are not too much different to the computerized gait analysis machine. Pre-and post-treatment gait analysis need to use the same methods; either computerized or conventional analysis.

CONCLUSION

Conventional analysis can be used in the community gait analysis or in the clinical setting that would like to evaluate only spatial and temporal gait measurement. The normative data in eldery are not too much different to the computerized gait analysis. Pre-and post-treatment gait analysis need to use the same methods; either computerized or conventional analysis.

REFERENCES

1. Simoneau GG. Kinesiology of walking. In: Neumann DA., ed. Kinesiology of the musculoskeletal system. Foundation for physical Rehabilitation.St.Louis:Mosby,2002:523-69
2. Pease WS, Quesada PM. Kinematics and kinetics of gait. In: Braddom RL, ed. Physical medicine and rehabilitation.Philadelphia:WB Saunders,1996:83-103.
3. Lehmann JF,De Lateur BJ. Gait analysis: diagnosis and management. In: Kottke FJ, Lehmann JF. Krusen's Handbook of physical medicine and rehabilitation. 4 th. ed. Philadelphia: WB. Saunder,1990:108-25.
4. Nijders AH. Van de Warrenburg BP, Giladi N, Bloem BR. Neurological gait-disorders in elderly people: clinical approach and classification. Lancet Neurol 2007;6:63-74.

UNILATERAL INTERMITTENT CLAUDICATION AFFECTS JOINT KINEMATICS DURING GAIT

Jennifer M Yentes[1], Panagiotis Koutakis[1], Sara A Myers[1], Jeffrey P Kaipust[1],
Iraklis Pipinos[2, 3], Jason M Johanning[2, 3], Nicholas Stergiou[1, 4]

[1]Nebraska Biomechanics Core Facility, University of Nebraska at Omaha, Omaha, NE
[2]Department of Surgery, University of Nebraska Medical Center, Omaha, NE
[3]Veterans Affairs Medical Center, Omaha, NE
[4]College of Public Health, University of Nebraska Medical Center, Omaha, NE
Email: nstergiou@mail.unomaha.edu Web: http://biomech.unomaha.edu/

INTRODUCTION

Peripheral arterial disease (PAD) affects over eight million people in the US and is a manifestation of atherosclerosis leading to decreased blood flow to the legs. The result is ischemic pain (claudication) that is induced by physical activity and results in diminished ability to walk [3].

CLINICAL SIGNIFICANCE

Minimal research work has been done to determine the gait handicap of PAD patients. Recently we have found that patients with bilateral PAD have altered ground reaction forces, joint kinematics, and spatiotemporal parameters as compared to controls [1, 4]. The present investigation extends this work to the evaluation of kinematics in patients with unilateral PAD. We hypothesized that the kinematics of unilateral PAD patients would be different between the affected and the non-affected limbs and as compared to healthy matched controls. We also hypothesized that these differences would become worse after the onset of claudication.

METHODS

Thirteen patients with unilateral PAD (age: 61.69$\pm$10.53yrs, mass: 84.65$\pm$20.24kg) and eleven matched healthy controls (age: 66.27$\pm$9.22yrs, mass: 77.89$\pm$10.65kg) walked at their self-selected pace along a ten meter pathway while kinematic data were recorded with an 8-camera motion capture system (Motion Analysis, Santa Rosa, CA). Five trials were collected for each limb in the pain free (prior to the onset of claudication) and pain induced conditions. The pain free condition was acquired first and one minute rest periods were required between each trial. After completion of the pain free condition, pain was induced by having subjects walk on a treadmill at 10% grade and 0.67 m/s. At the onset of pain, the pain induced condition was collected with no rest between trials. The group means of specific spatial-temporal parameters (Table 1) and the range of motion (ROM) from the hip, knee, and ankle joint during stance from the PAD patients were subjected to a 2x2 fully repeated measures ANOVA (Limb x Condition). Independent *t*-tests were also used to compare the group means from both conditions and both limbs of the PAD patients with the healthy controls.

RESULTS

Results are presented in Table 1 below.

DISCUSSION

Regarding the spatio-temporal parameters, our results demonstrated that compared with healthy controls, unilateral PAD patients walk slower, with decreased cadence and stride length, and increased step width. Thus, our results are in agreement with previous studies [1, 2] and reveal a behavioral similarity for these parameters between unilateral and bilateral patients. In addition, they confirm that there is a definite alteration in the gait function of unilateral PAD patients even prior to the onset of claudication. However, claudication further decreased velocity, stride and step length, and increased ankle ROM, indicating that pain worsened the gait of these patients. Stride and step length also revealed some symmetry problems between the two limbs, which were not supported by the ROM results. Taken together our findings establish the need for more research to better describe the gait handicap of these patients using gait analysis.

REFERENCES

1. Chen et al. (2008). Journal of Biomechanics, 41:2506–2514.
2. McDermott et al. (2001). Journal of the American Geriatrics Society, 49:747-754.
3. Rosamond et al. (2007). Circulation, 115(5): e69-171.
4. Scott-Pandorf et al. (2007). Journal of Vascular Surgery, 46(3): 491-9.

ACKNOWLEDGEMENTS

Support that was provided by the American Geriatrics Society's Hartford Foundation Dennis W. Jahnigen Award and the Alexander S. Onassis Public Benefit Foundation.

Table 1: Spatio-temporal parameters and joint kinematics for the stance phase of walking.

	Pain free Non-Affected Limb	Pain free Affected Limb	Pain induced Non-Affected Limb	Pain induced Affected Limb	Control Pain Free
			SPATIAL-TEMPORAL		
Gait velocity (m/s)	1.07±0.15*	1.04±0.16*	1.01±0.15*†	1.02±0.14*†	1.37± 0.15
Stride length (m)	1.30±0.12*§	1.27±0.13*	1.22±0.11*†§	1.24±0.10*†	1.51±0.10
Step length (m)	0.65±0.07§	0.60±0.06*	0.59±0.06*†§	0.60±0.05*†	0.68±0.05
Cadence (steps/min)	98.65±7.4*	98.68±8.3*	99.21±8.29*	100.06±7.7*	110.09±9.09
Step width (m)	0.14± 0.04*	0.14±0.03*	0.15±0.07*	0.14±0.04*	0.13 ± 0.03
			JOINT KINEMATICS		
Ankle ROM (degrees)	17.33±2.56	15.98±4.43	19.08±2.9*†	17.30±4.64†	16.22±3.53
Knee ROM (degrees)	8.22±3.60	6.90±3.50*	9.41±5.00	7.47±4.09*	10.24±3.27
Hip ROM (degrees)	35.14±4.2*	34.19±2.84*	35.13±5.7*	33.36±5.04*	39.42±3.12

Note: * $p < 0.05$, significant differences with healthy controls
† $p < 0.05$, significant main effect for pain
§ $p < 0.05$, significant main effect for limb
No significant interactions